Technical

Report

Annual Report
A-B2299-3

SPACE RELATED BIOLOGICAL AND INSTRUMENTATION STUDIES

by

R. J. Gibson
R. M. Goodman

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March 1968 to March 1969

Prepared for

NATIONAL AERONAUTICS AND SPACE ADMINISTRATION

Contract No. NSR-39-005-018

THE FRANKLIN INSTITUTE RESEARCH LABORATORIES
BENJAMIN FRANKLIN PARKWAY • PHILADELPHIA, PENNA. 19103

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FOREWORD

Experimental and research effort has continued on the low-cost, high reliability, multichannel implantable telemeter. Basic subcarrier oscillator designs have been completed and sufficient tests run to indicate their acceptability. A first four-channel fm/fm implant is in process of fabrication and is described.

Studies of ion concentration transducers have been carried out, and the experimental development of related pH sensors and reference electrodes is reported.

A *first* monograph resulting from our Managed Energy Terrella (MET) work is included in this report. It is bound separately as Volume II for the convenience of the reader. Other monographs are expected to follow.

Cooperation and exchange of information with other workers has continued throughout the period covered by this report and details are given.

It is with pleasure that we add the observation here that much, if not all, of the research thinking and development efforts supported under this contract are of direct and useful interest (and application) to the community of biological workers. This includes those pursuing studies in biological rhythms and those directly related to the development and application of artificial assist devices and replacement organs for humans.

We hope that our efforts related to the description of biological experiments (MET) and of special techniques and methods will be of help to those interested in the interactions between living systems and their environment. Again, such a study area is of equal interest to those biologists concerned with space problems as it is to those dealing with interactions on earth.

ACKNOWLEDGEMENTS

The authors acknowledge contributions to this report via the skill, ingenuity and suggestions of Mr. L. Aruffo, Mr. A. Marmarou, Mr. J. Price and Mrs. T. Webster.

TABLE OF CONTENTS

LIST OF ILLUSTRATIONS

1.0 MULTICHANNEL TELEMETRY

Our effort relating to the development of implantable multichannel telemeters for use in chronic studies has been under way for some time. Early work has been reported[1-1,1-2,1-3] elsewhere. Sufficient data and experience are now in hand so that a practical and useful reduction to practice is possible. Our work in this regard is reported here.

1.1 SUBCARRIER OSCILLATORS

The Wien-Bridge subcarrier oscillator (SCO) illustrated in Figure 1.1-1 has been developed as our basic design.

Detailed studies of stability as a function of ambient temperature and applied voltage have been carried out. Figure 1.1-2 illustrates a specific SCO designed for a center frequency (f_c) of 1.7 KHz. Its performance with temperature and voltage is illustrated by Figure 1.1-3. R_x in Figure 1.1-2 represents the frequency-varying element and would, in a practical situation, be replaced with a suitable transducing element.

The circuit was subjected to a supply voltage variation of 1.345 to 1.330 volts. These limits were selected since we plan to use the mercury cell as a prime power source and because our measurements and observations indicate that about 90% of the useful life of the cell exhibits an output voltage in the cited range.

Referring again to Figure 1.1-3 we determine the following:

(1) for R_x = 115.6 KΩ

V = 1.345; $25 \leq T \leq 45°C$; $\bar{f}$ = 1613 Hz

V = 1.330; $25 \leq T \leq 45°C$; $\bar{f}$ = 1617 Hz

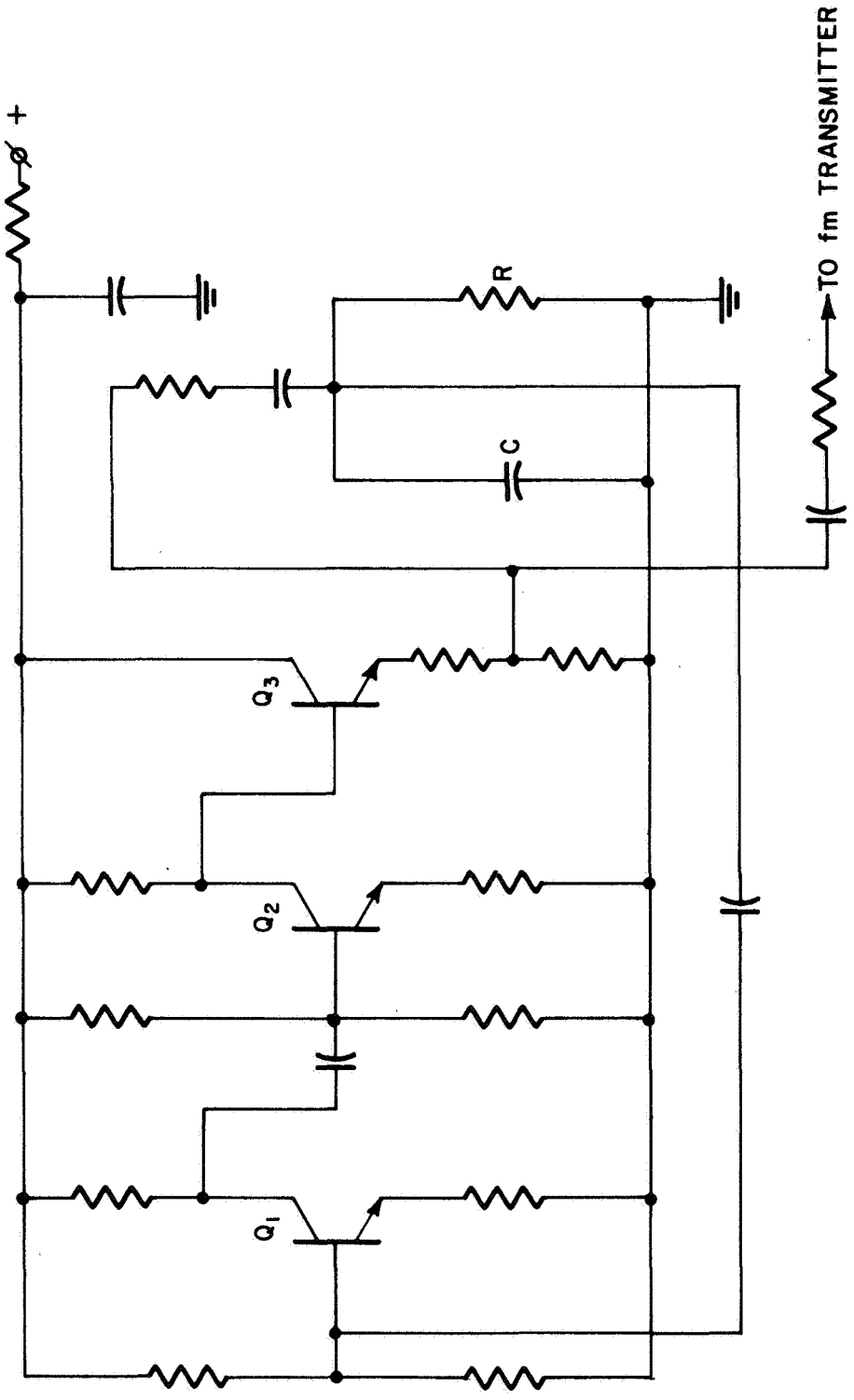

Figure 1.1-1. Typical Subcarrier Oscillator

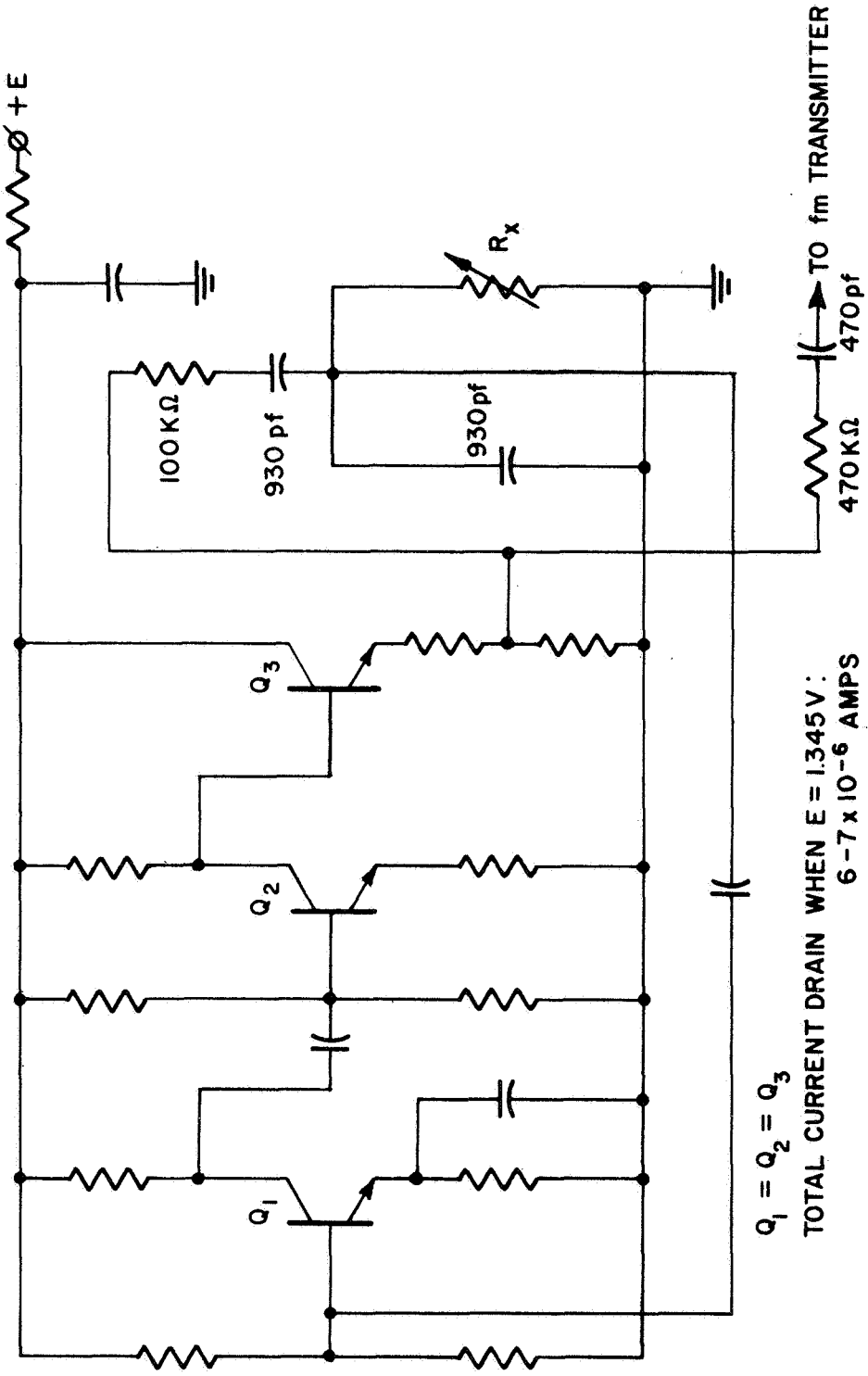

Figure 1.1-2. Subcarrier Oscillator, f_c = 1.7 KHz

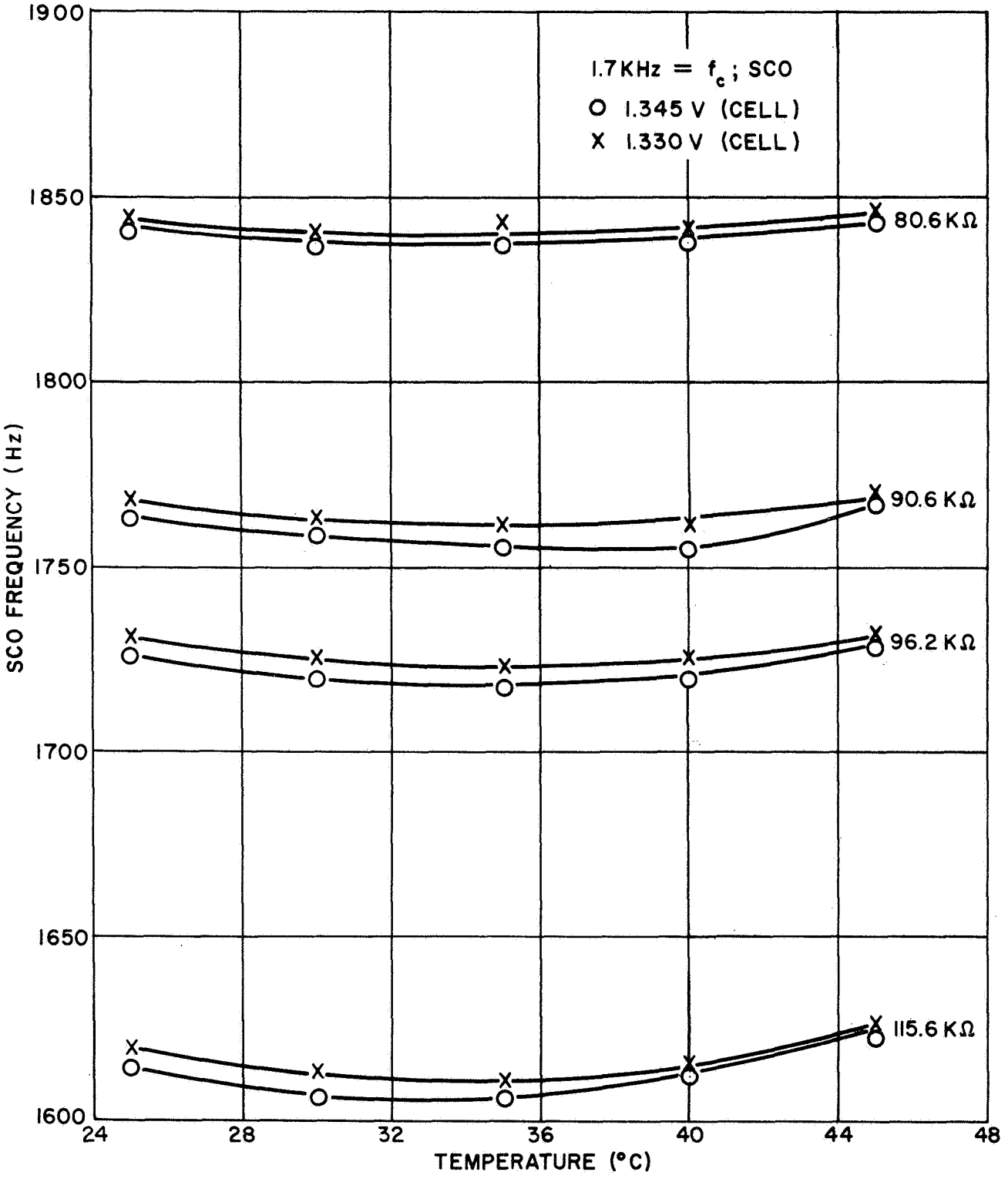

Figure 1.1-3 SCO, Frequency vs. Temperature

Taking the mean frequency as 1615 Hz, maximum deviations from the mean value occur at

$T = 45°C$, $V = 1.330$: $\Delta f = +9$ Hz (+ 0.56%)

$T = 35°C$, $V = 1.345$: $\Delta f = -9$ Hz (- 0.56%)

(2) For $R_x = 96.2$ KΩ

$V = 1.345$; $25 \leq T \leq 45°C$; $\bar{f} = 1723$ Hz

$V = 1.330$; $25 \leq T \leq 45°C$; $\bar{f} = 1728$ Hz

Taking the mean frequency as 1725.5 Hz, maximum deviations from the mean value occur at

$T = 45°C$; $V = 1.330$: $\Delta f = +6.5$ Hz (+ 0.38%)

$T = 35°C$; $V = 1.345$: $\Delta f = -7.5$ Hz (- 0.43%)

(3) For $R_x = 90.6$ KΩ

$V = 1.345$; $25 \leq T \leq 45°C$; $\bar{f} = 1760.4$ Hz

$V = 1.330$; $25 \leq T \leq 45°C$; $\bar{f} = 1765.2$ Hz

Taking the mean frequency as 1762.8 Hz, maximum deviations from this mean value occur at

$T = 45°C$; $V = 1.330$: $\Delta f = +7.2$ Hz (+ 0.41%)

$T = 40°C$; $V = 1.345$: $\Delta f = -7.8$ Hz (- 0.44%)

(4) For $R_x = 80.6$ KΩ

$V = 1.345$; $25 \leq T \leq 45°C$; $\bar{f} = 1840$ Hz

$V = 1.330$; $25 \leq T \leq 45°C$; $\bar{f} = 1843$ Hz

Taking the mean frequency as 1841.5 Hz, maximum deviations from this mean value occur at

$T = 45°C$; $V = 1.330$: $\Delta f = +6.5$ Hz (+ 0.35%)

$T = 30°C$; $V = 1.345$: $\Delta f = 3.5$ Hz (-0.13%)

We note that the range of 25 to 45°C represents useful extremes for biological temperatures in most homeothermic subjects. When performance is considered in the range of 30 to 40°C (similar to Mark IV implant units), the acceptable performance reported above improves further.

Now, referring to Figure 1.1-4, it is clear that computed values

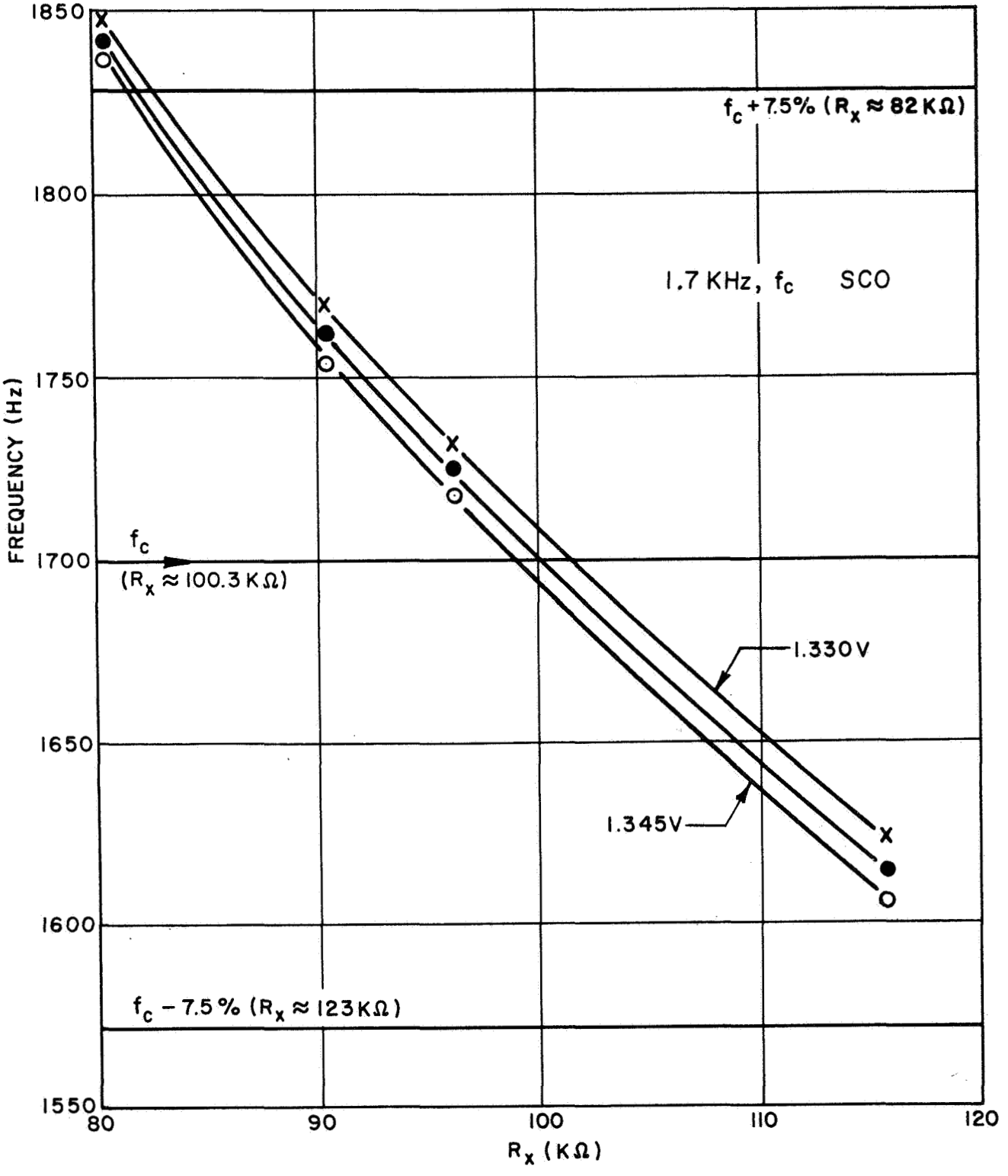

Figure 1.1-4 SCO, Frequency vs. Rx

have given us a predicted f_c = 1.7 KHz, (in the physical circuit R_x = 100.3 KΩ). And for a range of 82 KΩ≤R_x≤123 KΩ a deviation of ∿ 7-1/2% in SCO, f_c will be accomplished. Thus assuming we wish to sense a 15°C temperature range (say, 30 to 45°C), then

let R_x = 100 KΩ at 37.5°C
= 82 KΩ at 30°C
= 123 KΩ at 45°C

Substitution of a typical thermistor (300 KΩ@ 25°C) in parallel with a 270 KΩ resistor for R_x will obtain ca. 82 KΩ at 45°C and ca 126 KΩ at 30°C. As a matter of interest it can be shown that the response to sensed temperature will be essentially linear over the desired range.

Using the parallel circuit described above will produce the following least counts for the SCO's listed:

f_c	least count (°C)
5.4 KHz	0.019
3.9 KHz	0.026
3.0 KHz	0.033
2.3 KHz	0.044
1.7 KHz	0.059
1.3 KHz	0.078
0.96 KHz	0.104
0.73 KHz	0.137

Substitution of a resistive semiconductor in place of R_x is clearly useful as demonstrated above. We can also accomplish control of R_x by insertion of a suitable field-emission-transistor in its place. Thus, the circuit of Figure 1.1-2 becomes that of Figure 1.1-5. Fortunately, FET's of choice can be obtained in "T-Pac" form and are small enough for inclusion in lumped-parameter constructs for multichannel implants. Signals on the order of 1.5 x 10^{-3}V (pk-pk) at the FET gate input will produce deviations in f_c of ca. 2%.

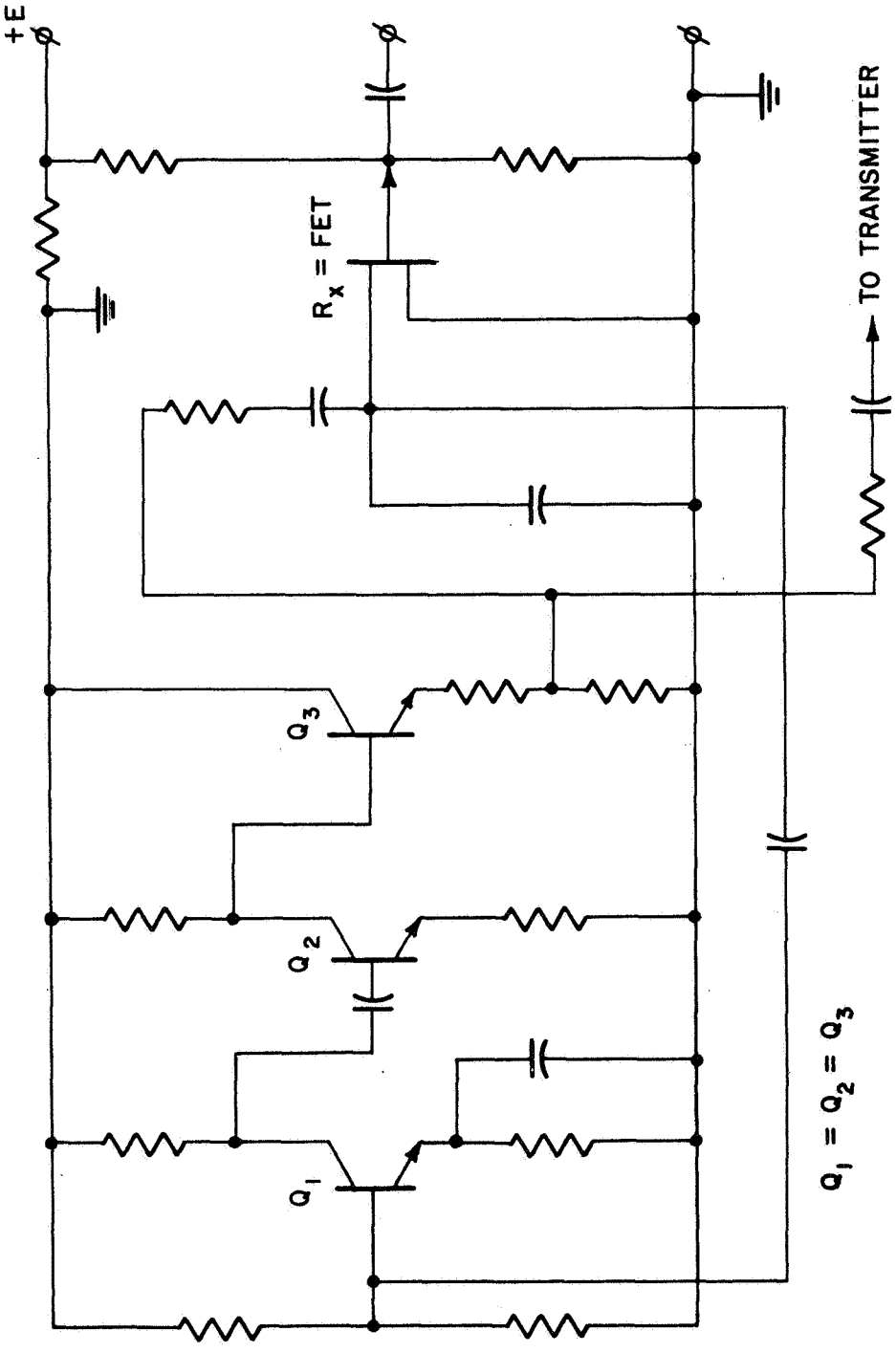

Figure 1.1-5. SCO with FET Deviation Control

1.2 SIGNAL CONDITIONING

We have been concerned with the conditioning of sensed information to produce stable and repeatable related deviations in the f_c of the SCO. Consideration of possible biological parameters of interest and the characteristics of appropriate sensors is pertinent here.

Temperature via thermistor sensing; sensor output is in terms of variable resistance. This presents no problems. See Section 1.1 above for details. It is useful to require, however, that the complete implanted system have the capability for response times of < 10 seconds (63%). It will then be possible to detect with satisfactory accuracy, temperature variations with frequency components more rapid than 1 cycle/minute.

Proper selection of a suitable thermistor and care in its incorporation (mechanically) into the system, will assure the attainment of a 10-second response.

ECG, via stainless steel loop electrodes; sensor (electrodes) output is in terms of a varying voltage from a source impedance of < 5 KΩ and which contains no dc component of interest. Acceptable ac frequency bandpass is ca. 0.5 Hz to ca. 200 Hz. Available signal levels are expected to be in the range of 0.05 to 2 x 10^{-3} volts.

Adequate signal conditioning can be accomplished via the amplifier illustrated in Fig. 1.2-1. This simple circuit requires only a few microamperes of current for its operation. It has a response of < 0.3 Hz to > 1 KHz. Its input resistance is on the order of 200 KΩ.

pH via FIRL pH electrode (See Section 2.7), sensor output is in terms of ca. $\mp$ 60 mv/pH unit and ca. 300 mv at pH = 7.0. Preliminary data indicates that no more than 10^{-7} amperes/cm^2 of current should be drained from the pH electrode. This will require a $10^8\Omega$ resistor in series with the electrode, feeding a $10^6\Omega$

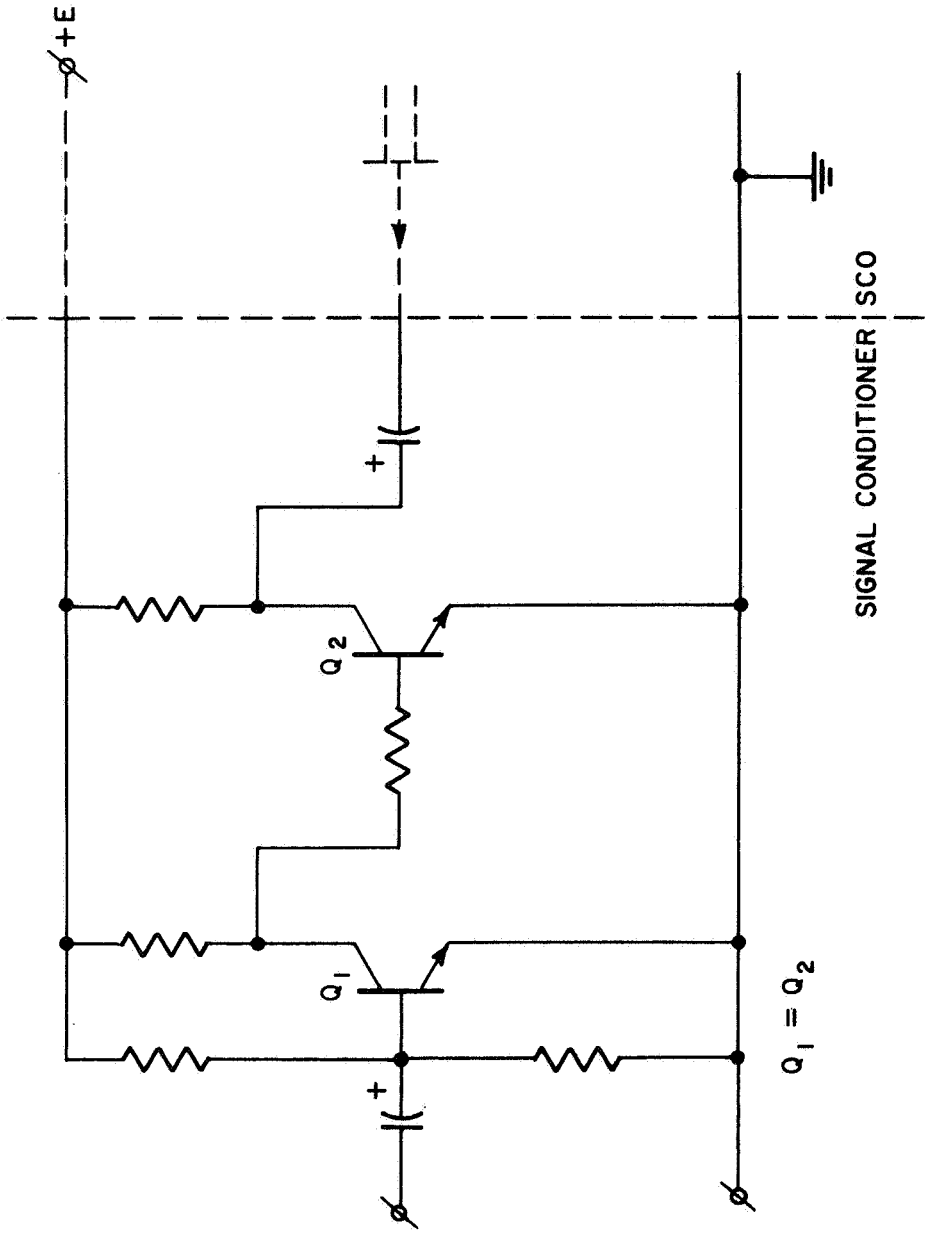

Figure 1.2-1. ECG Signal Conditioner

(Z_{in}) amplifier of low gain (1 to 2). The output from such a conditioning stage will be adequate to drive the SCO of Figure 1.1-5 on a *dc* basis. There are, of course, problems related to the use of FET's. Primarily these are related to their temperature sensitivity. Approaches to FET bias stabilization with temperature were discussed in our last annual report (ibid). The techniques noted are expected to be satisfactory.

Blood Flow via the crossed field transducer; sensor output is in terms of μv/ml/minute of flow. We have been fortunate in establishing no cost cooperation and information exchange with Dr. T. Reich of the Institute of Rehabilitation Medicine, New York University Medical Center. Of specific pertinence here is the fact that Dr. Reich and his colleagues have been actively developing a permanent-magnet implantable flow meter. It requires *no* activating power for its operation. Its output is on the order of 1/2 to 1 μv/ml/minute of blood flow. We do not yet have a really adequate signal conditioning circuit for this transducer. This statement is particularly true in the consideration of low flow rates. The requirements include the handling of a bandpass from dc to > 200 Hz and the provision of a stable baseline (input = 0 ml/min). It is our hope that continued cooperation between our Institutions will result in an enhanced transducer output which will permit us to make a practical attack on the signal conditioning problem.

Blood Pressure via the piezo-resistive and piezojunction sensors; signal output can be expressed in terms of ΔV or ΔR corresponding to a ΔP. Investigation of the, PITRAN (Stow Labs) pressure sensor has just begun. There exists a reasonable possibility that this sensor can be incorporated almost directly in place of R_x (Figure 1.1-2). If this approach proves feasible, virtually no signal conditioning will be required. If this is not the case, a unique pulse scheme has been evolved to permit

PITRAN operation with low duty cycle and relatively high-power pulses. This latter approach, while somewhat complicated, will sharply reduce overall power requirements for the pressure telemetry channel.

Our colleagues at NYU have, in active development, a pressure sensitive, diffused piezo-resistive bridge sensor. Its impedance (activating power load) is ca. $10^3\Omega$. If we elect to use it instead of the PITRAN, we shall apply our low duty cycle, pulse-drive scheme.

EMG via stainless steel electrodes; signal output similar to that range described for ECG; frequency bandpass > 1 KHz. Signal conditioning here will be similar to that described for ECG.

EEG We have not considered this parameter to date since we see no purpose in using an implantable system to acquire EEG data. However, if a need should arise to use the Mark V development for EEG data, we anticipate no problems with signal conditioning.

dc Potential, via membrane protected Ag-AgCl electrodes, signal output from 0.05 to several millivolts. Active attack on conditioning of signals from such electrodes awaits more sensor data.

1.3 THE MARK V, FOUR CHANNEL IMPLANT

As indicated in our previous report (ibid), a first multichannel implant will consist of three temperaturd channels and one for ECG.

The complete circuit for this device is shown in Figure 1.3-1. It consists of three SCO's, each of which is deviated by the variation in resistance value of a related thermistor. The thermistors are 0.035" in diameter and will be able to provide a highly localized temperature sensing capability. Based on the work described in previous paragraphs, we anticipate the following least counts in temperature sensing:

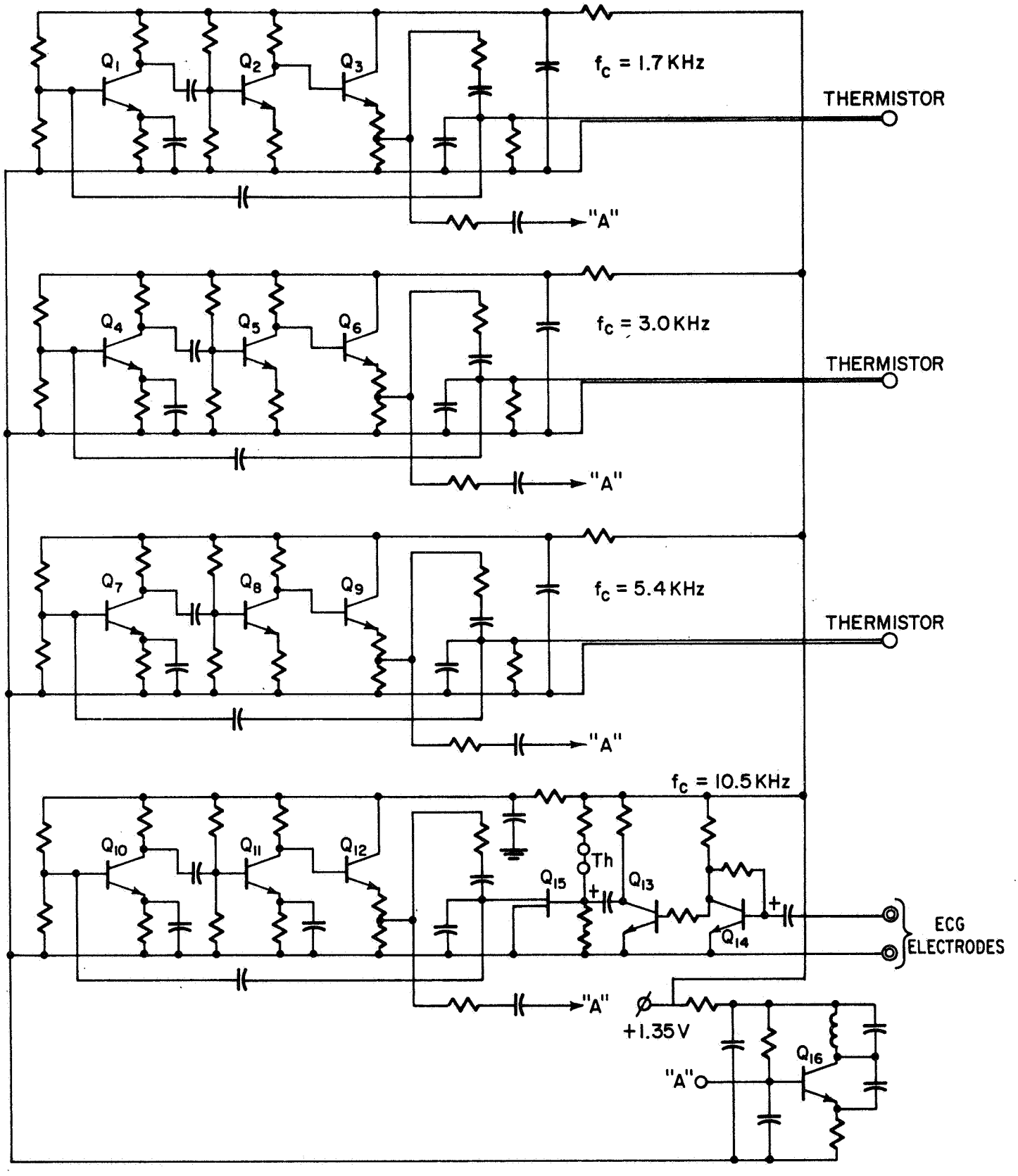

Figure 1.3-1. Mark V, Four-Channel Implant Circuit

Channel 1: 0.059°C

2: 0.033°C

3: 0.019°C

The ECG electrodes will be stainless steel loops attached to stainless, composite, multistrand, helically wound leads to minimize breakage. If it should prove desirable at a later date, we will implant an externally operated magnetic mercury switch (volume: 4×10^{-4} in^3) to conserve battery power.

The circuit has been reduced to a layout and wiring diagram as illustrated in Fig. 1.3-2. This figure is illustrative of the technique we have developed as an aid in the fabrication of these dense circuits. The first step of the process is an actual layout 10X actual size. The resulting pictorial is then reduced in two photographic steps to 5X size and to 1X size. The final positive prints are made on Kodak Estar film which is of preselected thickness; the film becomes the actual supporting chassis in the telemeter. Figure 1.3-3 is a photograph of the three full size (1X) chasses designed for the Mark V. These chasses fit into a supporting structure which also acts as the transmitter-oscillator tank coil support. The arrangement can be seen in the photograph of a preliminary test-unit layout, see Figure 1.3-4. In the same photograph it can be seen that the unit is to be contained in an "inner" shell, the two halves of which are shown. This shell, cast of syntactic epoxy is extremely rugged and of very low density. It permits us to avoid coating the circuitry with impermeable paraffin and reduces transmitter "pulling" when implanted. This latter observation will be discussed in more detail in Section 1.6 of this report.

The entire implant, including its power cell and inner container, is expected to weigh < 4 grams. Continuous operation of 4 to 5 months is anticipated in this size. When the unit is implanted in animals which are larger than small rats, modest increase in physical size will permit the use of mercury cells with performance capabilities of a continuous year or longer.

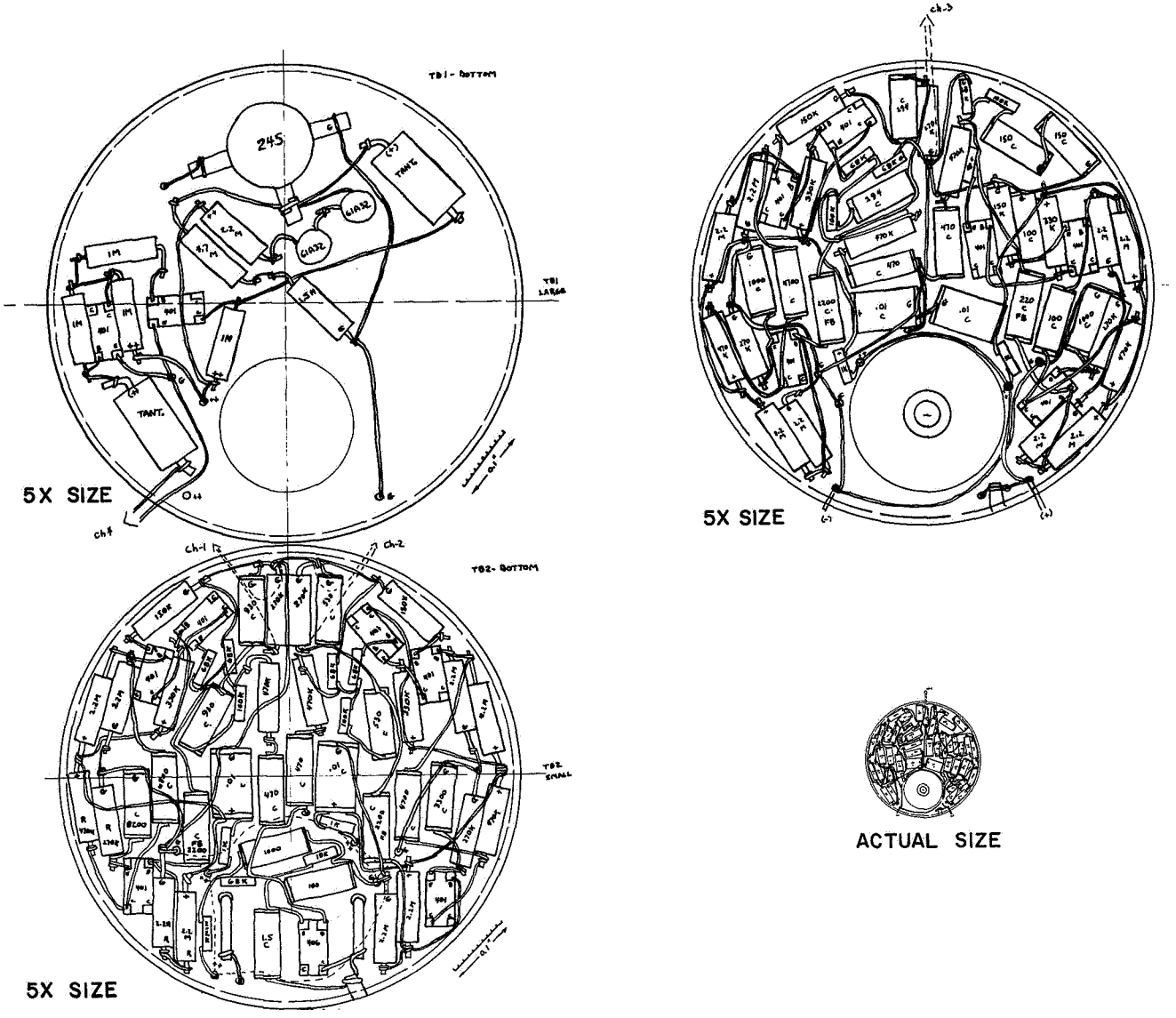

Figure 1.3-2. Four-Channel Wiring and Component Layout

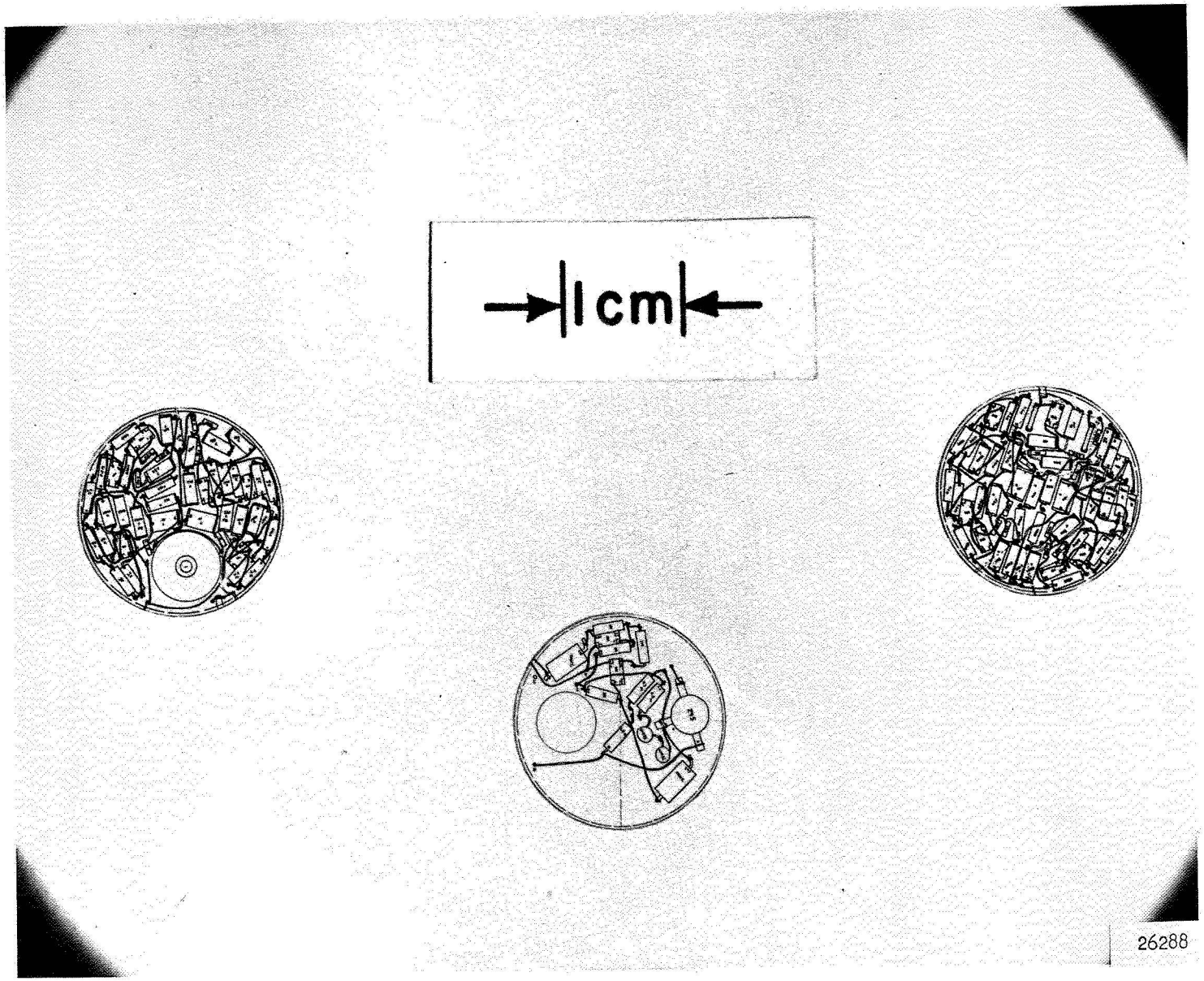

Figure 1.3-3. Chasses Set, Four-Channel Telemeter

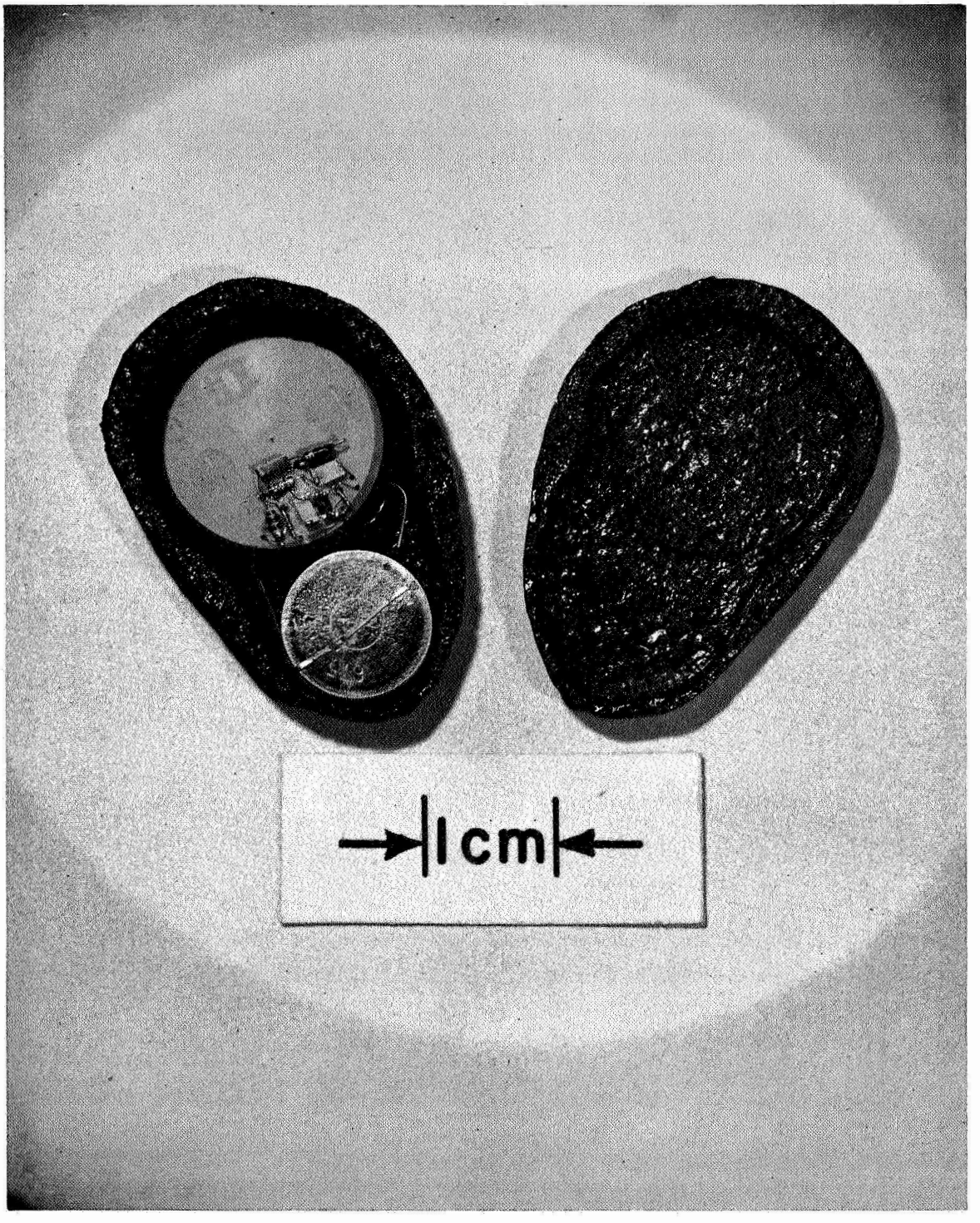

Figure 1.3-4. MkV Test Unit Layout, Preliminary

Almost all components are now in hand for this unit, and construction is expected to initiate in March 1969. At the same time, a demodulator "station" for four-channel data will be fabricated. This station will be designed so that it can be economically expanded for eight-channel operation.

1.4 A NEW TWO-CHANNEL IMPLANTABLE TELEMETER

In view of an increasing interest among rhythm biologists in the acquisition of ECG-complex and temperature data from small animals, we have designed a practical implant for the purpose.

Since only two data channels are involved, it was not necessary to use continuous transmission and sine-wave subcarrier oscillators. We were able to apply the complementary SCO developed with NASA support and reported at the Stockholm International Conference (ibid). We were also able to utilize the new experience acquired in the application of the resistive field emission transistor. The net result has been the development of an extremely low-power pulsing, fm/fm implant able to transmit simultaneously the ECG-complex and temperature.

The developed circuit is shown in Figure 1.4-1. Total current drain is 13×10^{-6} amperes. Operational life with the Y1404 mercury cell is > 5 months. Life with the Y1407 is expected to exceed two years. The unit will transmit usefully for distances of 1 to 2 meters when immersed in 2 liters of physiological saline. Generally, actual implant conditions do not present as severe an absorption problem as "solid" saline solution and the useful range is expected to be greater.

Based on earlier work, we have stabilized the pulsing subcarrier oscillator to minimize error introduced by temperature differentials between the circuitry and the remote thermistor sensor. Typical performance with temperature and cell voltage is illustrated in Figure 1.4-2. Figure 1.4-3 indicates unit response to temperature when a sensing thermistor is emplaced.

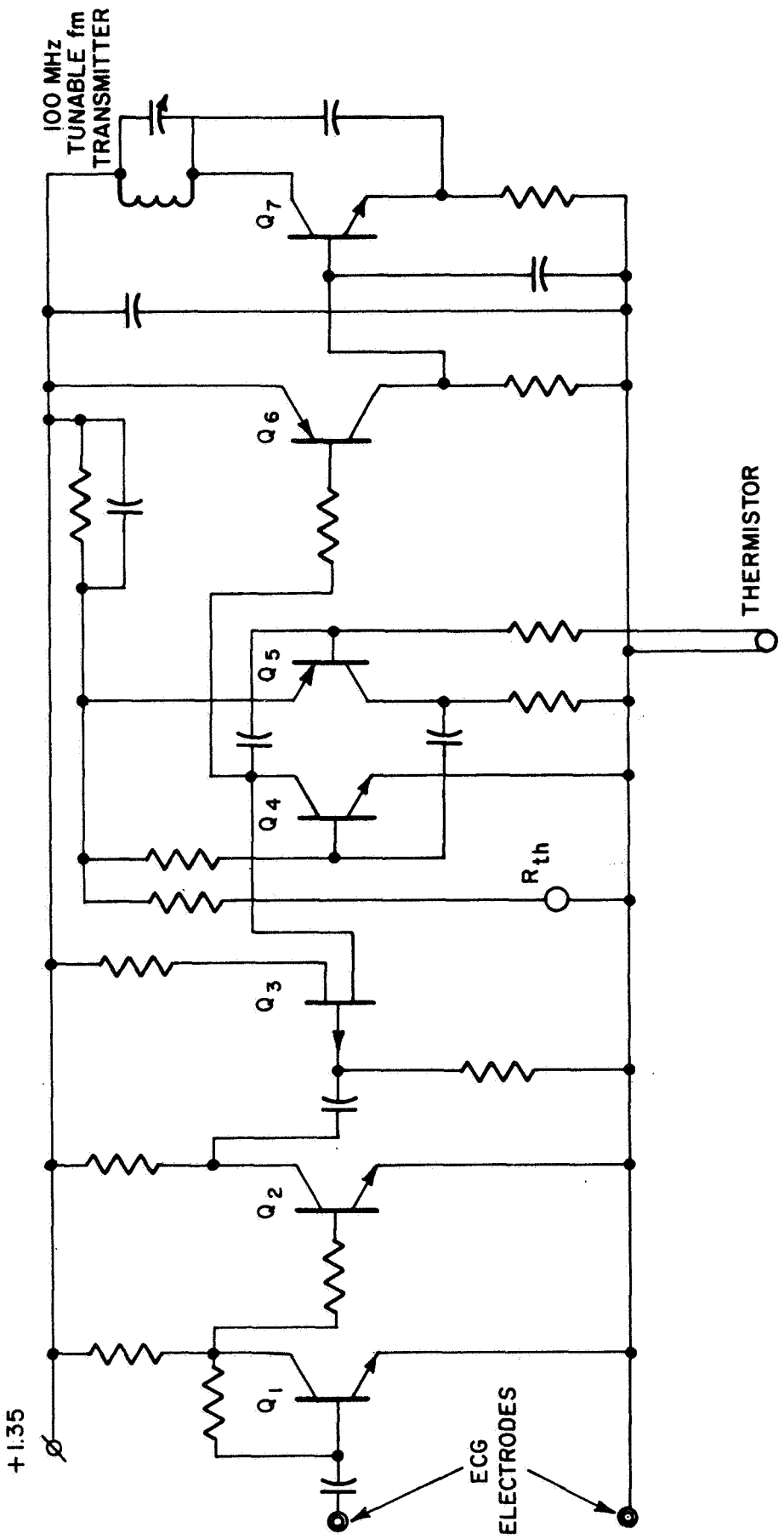

Figure 1.4-1. Dual Channel, fm/fm Implant (MkV-PCT)

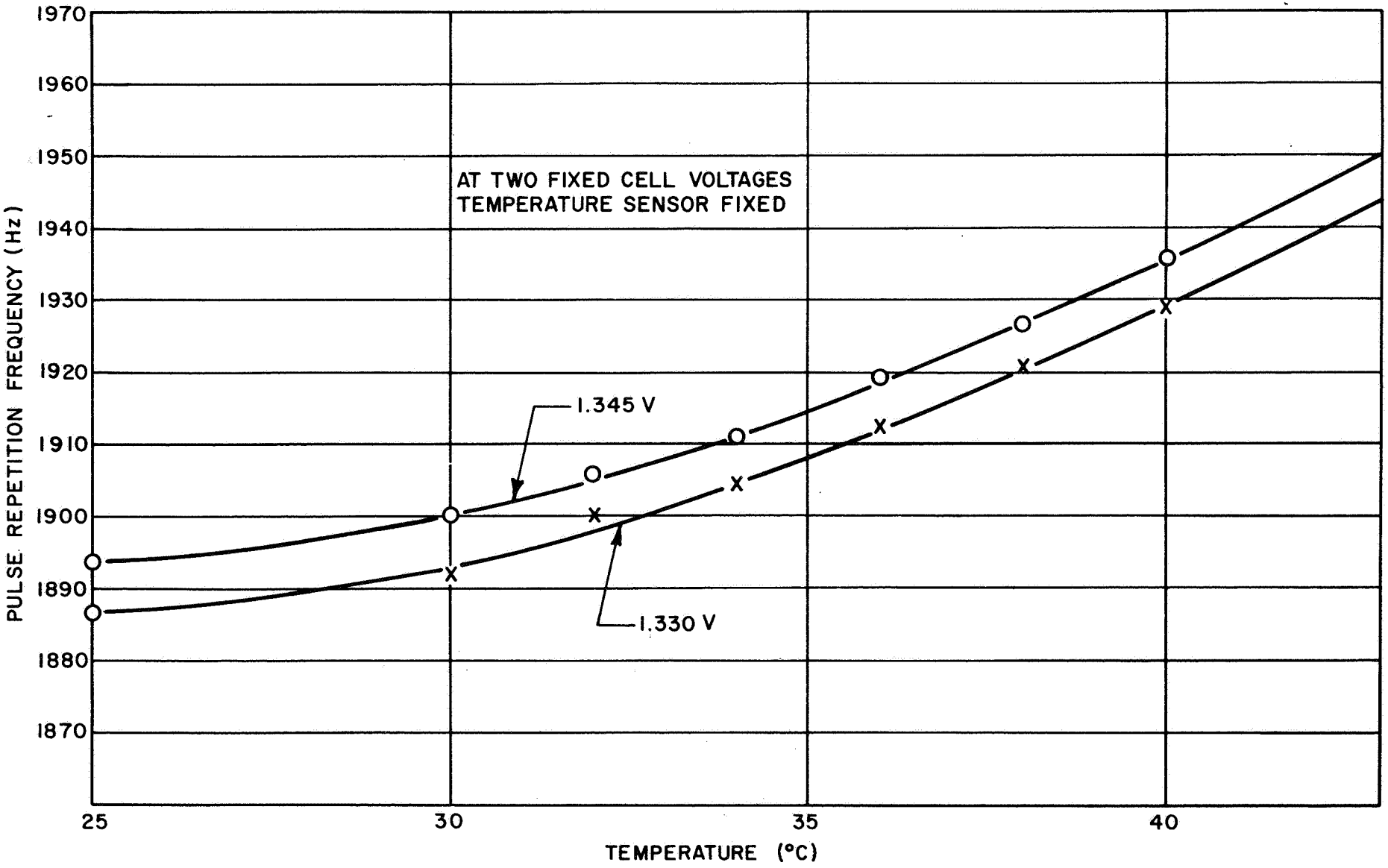

Figure 1.4-2. MkV-PCT Telemeter PRF Versus Temperature

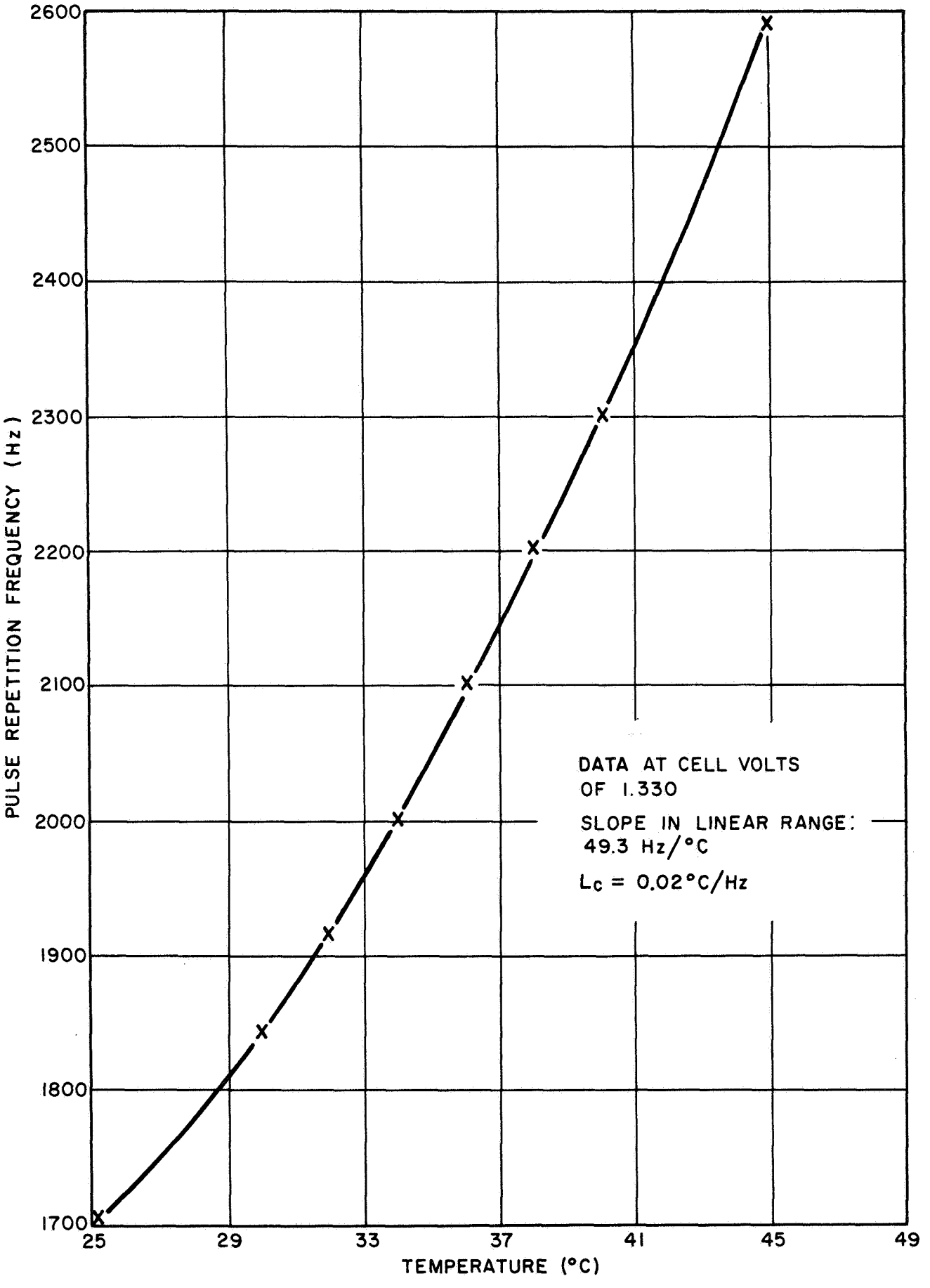

Figure 1.4-3. MkV-PCT Unit PRF Versus Temperature

From these curves, the following data are pertinent:

Anticipated Drift Data for Range of 1.345 to 1.330 Volts

At T = °C.	Drift in Hz	Equiv. Temp. Error, °C.
25	7	- 0.14
30	7	- 0.14
32	7	- 0.14
34	6-1/2	- 0.13
36	6-1/2	
38	6-1/2	
40	6-1/2	
42	6-1/2	

Since useful cell life is expected to run from 5 months to more than two years depending on cell selected, one may expect monthly drift figures (arising from cell voltage drop only) to be:

Cell	Drift/Month
Y1404	- 0.028°C
Y1407	- 0.006°C

We may also use these curves to estimate error introduced by a temperature differential between the telemeter circuity and the thermistor sensor. If one assumes the temperature differentials listed below, related data errors are expected tc be as indicated.

Sensed Temp., °C.	T, °C.	Data Error in °C.
25	+ 1	0.02
30	± 1	± 0.04
32	± 1	± 0.06
34	± 1	+ 0.07 - 0.06
36	± 1	± 0.08
38	± 1	+ 0.09 - 0.08
40	± 1	+ 0.10 - 0.09
42	± 1	± 0.10

Of course, if the sensing thermistor is located in close proximity to the telemeter circuitry these differential errors are further minimized.

A first experimental package was fabricated to permit us to run a series of simple tests. The unit is shown in Figures 1.4-4 and 1.4-5. While total unit weight (encapsulated) was ca. 5 grams, we are not satisfied with either the form factor or the wiring layout. A new form for the unit has been evolved which is expected to result in a smaller and much more rugged package. As soon as the new package is completed, we will arrange for its implantation. We believe this unit will be directly useful in the study of biological rhythms, pharmacologic toxicity evaluations, psycho-stimulus response studies, etc. The new wiring layout at 10X size is shown in Figure 1.4-6.

1.5 TELEMETER TRANSMISSION WHEN UNIT IS IMMERSED IN PHYSIOLOGICAL SALINE AND RELATED MATTERS

In the past, the Mark V telemeter development considered the use of fixed-tuned transmission circuitry on the basis that a spread of transmissions over the fm band could be obtained by selection of tuning capacitors. However, we reported earlier (ibid) that the use of Sylgard to protect circuit wiring produced a lowering of transmission frequency, a not surprising observation. Encapsulation of the entire unit with impermeable paraffin produced an additional drop in transmission frequency. Finally, immersion in saline produced a third drop in frequency. Of even greater importance was the fact that our transmitter input power had been optimized for lowest possible collector current consistent with stable operation. The transmitter circuit loading as indicated by the sequential frequency pulling noted above, forced the transmitter out of oscillation in some cases. We reviewed the situation and determined that:

(1) We could design a subminiature variable capacitor into the layout. This gave us the capability to set the transmitter frequency anywhere in the band without resorting to capacitor selection and stabilization.

(2) The circuit could be isolated with a syntactic epoxy which would introduce a minimum electrical load.

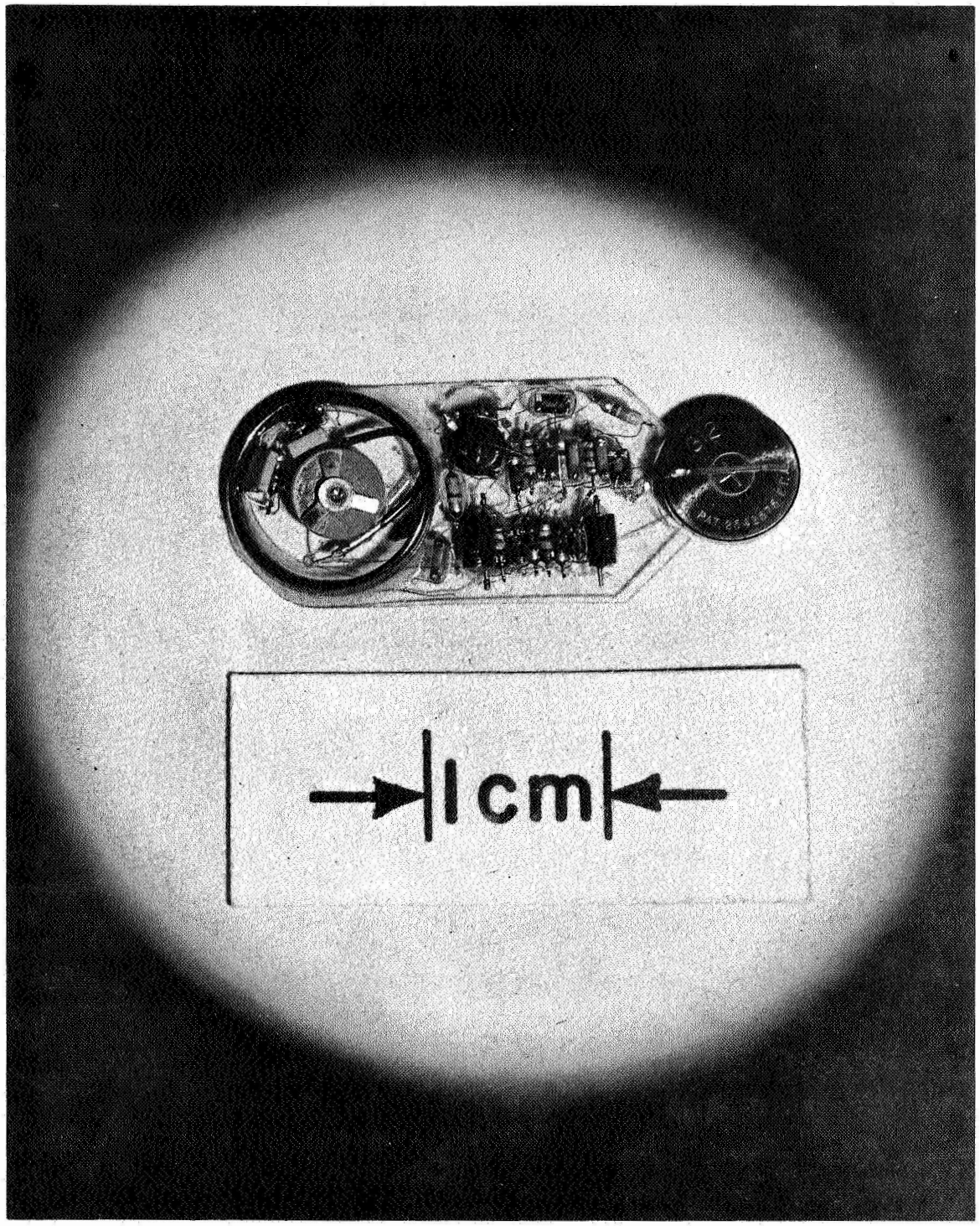

Figure 1.4-4. Experimental Dual Channel Unit, Internal View

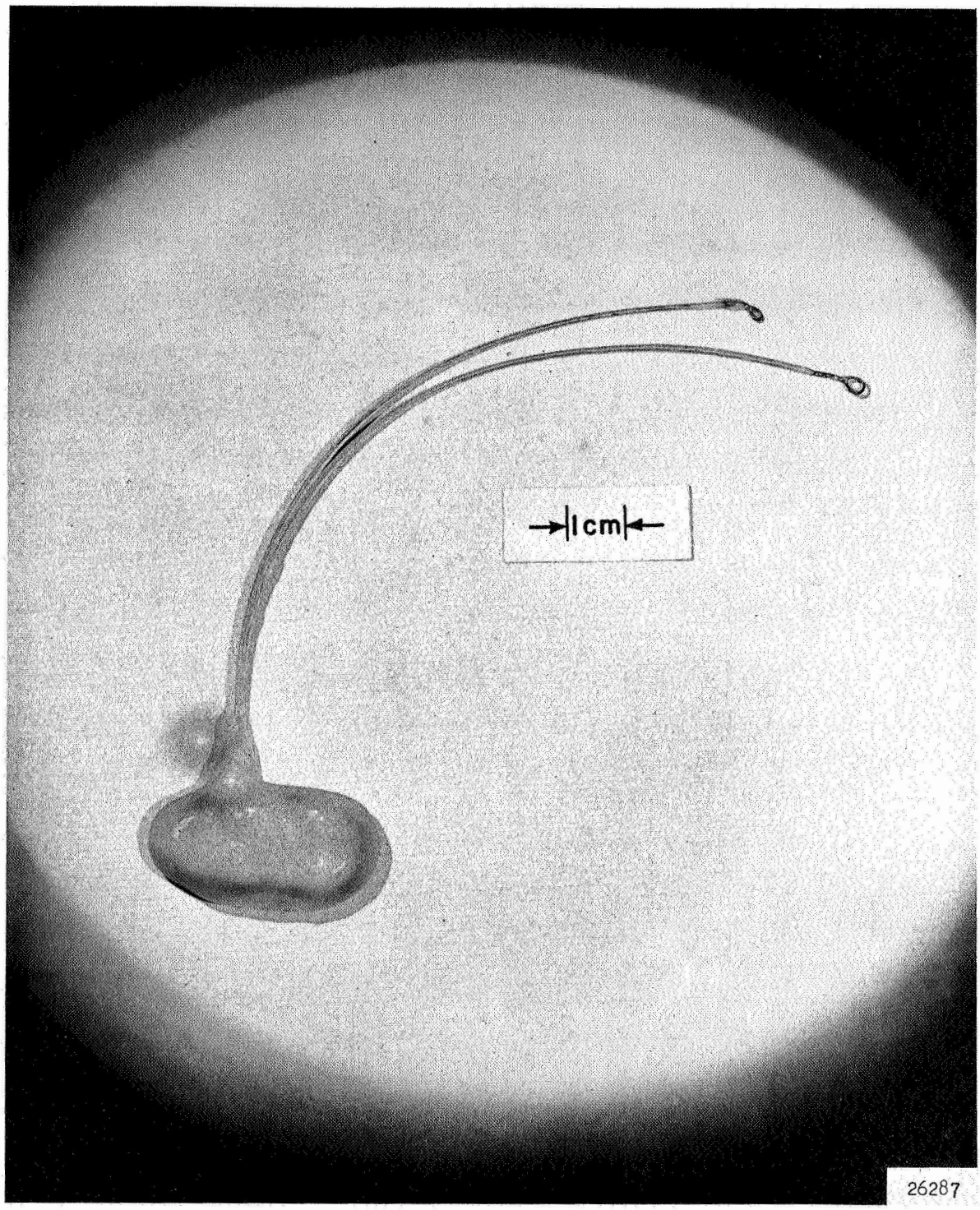

Figure 1.4-5. Dual Channel Telemeter, Preliminary Package

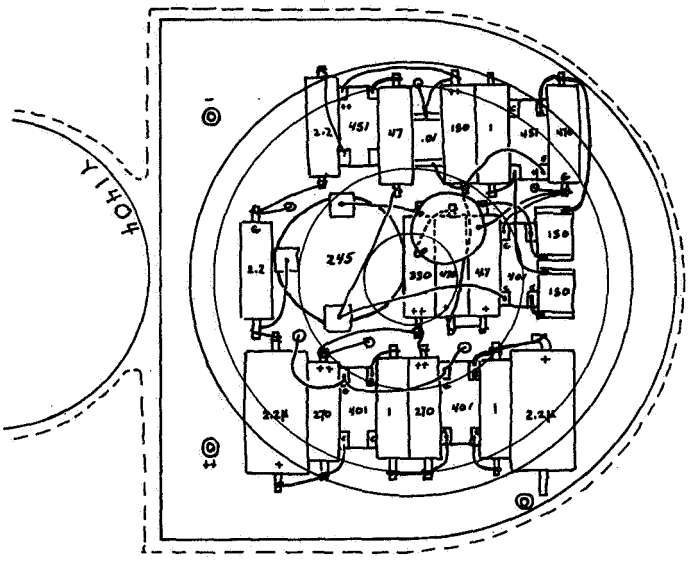

Figure 1.4-6. Dual Channel Telemeter, New Layout

(3) When the oscillator is protected by a syntactic epoxy barrier, an overcoating of impermeable paraffin has little loading effect on the transmitter.

(4) The transmitter collector current should be on the order of 40 μa to assure adequate transmission from the center of a right cylinder of normal saline, > 10 cm. in diameter and more than 10 cm in altitude. The frequency pulling effect from immersion in the saline solution amounted to 0.1 MHz or less.

The use of a syntactic epoxy barrier in this manner is of considerable interest here. Not only do we improve the telemeter electrical performance, but we have introduced an extremely rugged base to which we can anchor steel suture loops and which can be designed to anchor sensor umbilicals. Of additional importance is the fact that the use of an inner barrier obviates the necessity for circuit encapsulation with its attendent problems. Such circuit encapsulation makes repair extremely difficult. This latter point is a matter of some consequence since multichannel implants are not expected to be so inexpensive as to be casually disposable.

1.6 SPECIAL TOOLS AND TECHNIQUES

In the course of the development work discussed in the foregoing sections of this report, a number of techniques have been evolved which we hope will prove useful to other workers.

Reference to Figures 1.3-2 and 1.3-3 will illustrate the method used in laying out the extremely dense component packing for the multichannel implants. A first layout is made at 10X full size. This layout is accomplished by locating cardboard cutouts (accurately representative of individual components) in an optimum way. When the layout is complete, component outlines are sketched into the drawing and interconnecting wiring is indicated. The final drawing is an accurate representation of the telemeter component layout and essential interrelationships. This drawing (made on vellum) is then precisely reduced, by photographic methods, to 1X size on Kodak Estar film. The Estar film base is a rugged plastic which is stable and which is available in thicknesses

from about .0035" to .007". This process is both effective and economical. It eliminates any need for extremely complex and time consuming dimensioning and subsequent costly precision machining related to the chasses. Further, since components are normally fastened in place and interconnected with the aid of a microscope, the technician processing the chasses is working directly with a positive reference print.

The application of a syntactic epoxy as a structural material internal to the telemeter has proved to be a useful innovation. This material, Emerson and Cuming, Inc., Stycast 1090SI, has a specific gravity of 0.72 and a dielectric constant at 10^8Hz of ca. 1.8. Its dissipation factor is below .015. This material contains micron-sized, thin-walled bubbles made from silica. These bubbles are surrounded by an epoxy. The resulting material is structurally strong and light with excellent electrical characteristics. It can be cast with accuracy and when used as the "inner container" for the telemeter, it precludes the necessity for circuit immersion in encapsulant and provides a rugged platform for suture loops and umbilical anchorage.

In the course of working with densely packed circuits, we have shifted to the use of fine wire as "hookup" wire. Presently we are using .003 copper. This wire tins easily and very small quantities of solder are required. The use of gold-plated solder balls has become increasingly appropriate. In the past, soldering has been accomplished with small and light-weight irons. These are continuously operated at half-voltage until directly needed for processing a joint. At that time, the iron is brought to operating temperature with a foot-switch control. The tip of the iron is conical with a tip radius on the order of .005".

While our use of micro-tip soldering irons has been reasonably successful, we have initiated a small effort to investigate the value of hot gas as a soldering means. Figure 1.6-1 is a photograph of a number 25, stainless steel hyperdermic needle modified for hot gas use. We are using dry nitrogen to provide simultaneously an inert atmosphere and heat at the soldering joint. Gas flow rates are about 2.0 cc/min. This techniques shows some initial promise and will be studied and reported in more detail later.

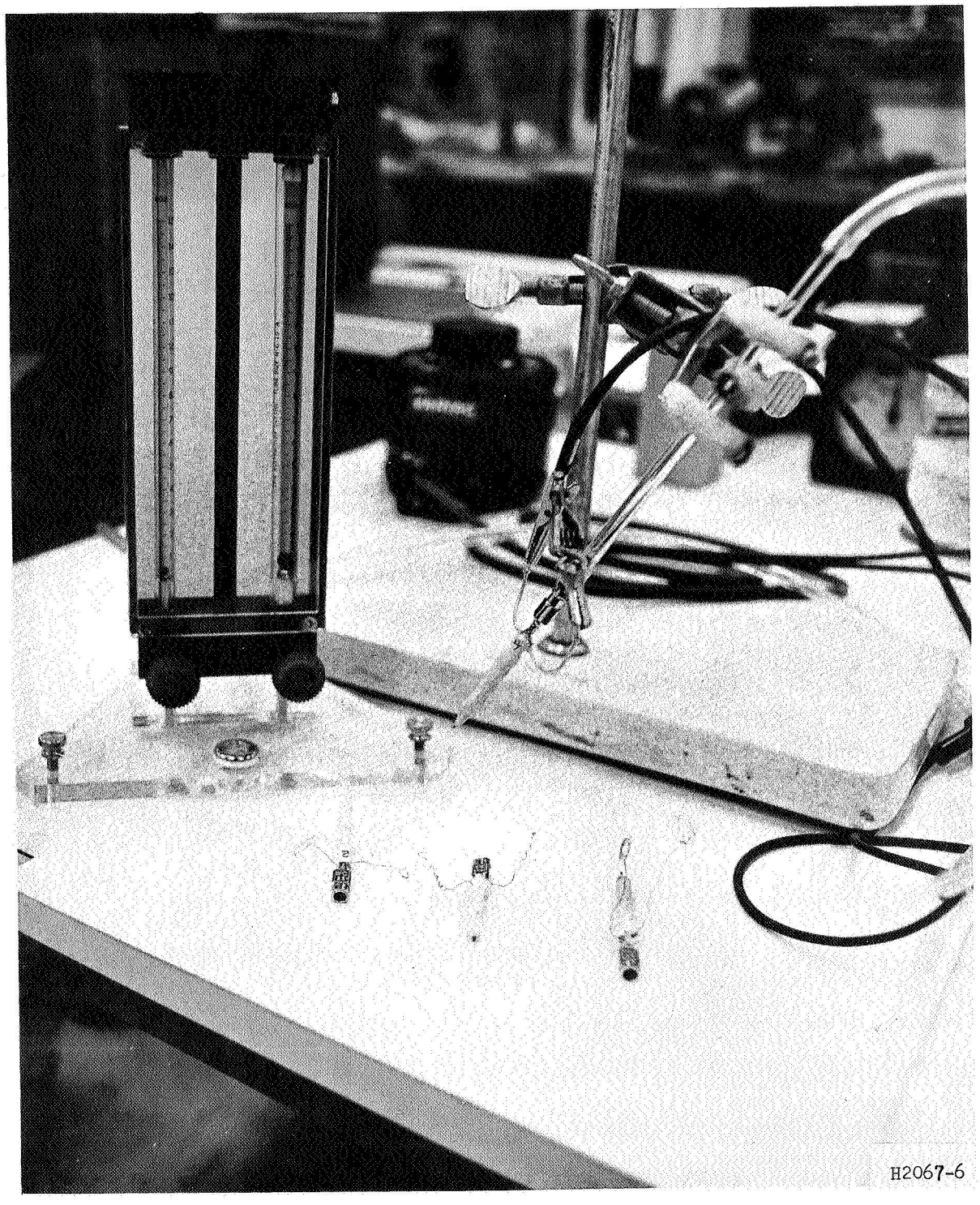

Figure 1.6-1. Inert Gas Soldering Tool

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1-2 Gibson, R. J., Jr., Goodman, R. M. and Isquith, I. R., "Space Related Biological and Instrumentation Studies, FIRL Report A-B2299-1, March 1966-March 1967.

1-3 Gibson, R. J. Jr. and Goodman, R. M., "Space Related Biological and Instrumentation Studies," FIRL Report A-B2299-2, March 1967-March 1968.

2.0 IMPLANTABLE TRANSDUCERS

2.1 ENCAPSULATION TECHNIQUES AND MATERIALS

Each sensor type presents a set of problems specific to itself. Thus in the following sections we will discuss encapsulation and materials as they may relate to the particular sensor involved.

2.2 TEMPERATURE SENSORS

In the past we have made considerable use of the Yellow Springs Instrument Company thermistors. These units have proved to be excellent in quality and performance. Their cost is modest. We find that there exists a desire (from the biologists) for more localized data; this is to say that many researchers wish to sense temperature at specific sites. It is our feeling that improved localization can be accomplished by us using smaller thermistors. To this end we have begun to apply units manufactured by the Victory Engineering Corporation. Excellent glass-coated thermistor units are available in diameters from .13 mm up through 1.1 mm.

These thermistors are directly applicable for internal telemeter compensation networks and as external temperature sensors. When used as external sensors, thermistor leads are fed through a small diameter polyethylene tube which is filled with flexible paraffin. The entire assembly is overcoated with a thin layer of flexible paraffin. This arrangement effectively seals the sensor against body fluids and at the same time provides an extremely flexible umbilical arrangement.

2.3 PRESSURE SENSORS

As indicated in Section 1.0 of this report, we are considering two possible sensors. The first is being developed by our colleagues at the New York University Medical Center, Institute for Rehabilitation Medicine[2-1]. Their pressure sensor is based on the use of a piezo-resistive Wheatstone

Bridge. The bridge is diffused into a 3 mm-diameter wafer of monocrystalline silicon. It is temperature compensated electrically with a set of "built-in" special resistors related to the bridge elements.

The bridge sensor is protected from body fluids by an extremely thin mica window (1-2 mg/cm^2) which is fused to the housing. This seal is hard-vacuum tight. Observations are presently in progress to assess if thrombogenicity is of consequence when the sensor is in contact with the blood. If a problem should occur, protective membranes of silastic and/or heparinized silastic will be evaluated. The sensor has a sensitivity of ca. 25 μv/mmHg/excitation volt.

The device presents an impedance of ca. 10^3 ohms to its excitation source. Thus at one volt dc, a milliwatt of power is required. To reduce this requirement we have evolved a pulsed-power scheme wherein the anticipated power duty-cycle will be 10% or less. This will mean 100 μw or less of sensor drive power.

The second sensor under consideration is the piezojunction unit called PITRAN. This interesting device is manufactured by the Stow Laboratory. It has already been used for biological application and is reported in the literature[2-2]. The PITRAN is commercially available. By its nature, the device appears to show a conversion capability (ΔP to ΔE) several orders of magnitude greater than other pressure-sensitive transducers. It is reported (personal communication with manufacturer) that after ca. 10^9 pressure cycles no mechanical or electrical shift in performance was detectable. It is also pertinent to note that sensitive diaphragm motion with pressure is measurable in low microinches. By its very stiffness we would anticipate that encapsulation of the sensor in suitable protective materials can be accomplished with no effect in the dc to low frequency characteristics of the device. Its normal response exceeds 100 KHz and with proper materials choice we will expect to attain uniform and useful response in the range of biological interest.

Because the PITRAN is essentially a transistor device, it offers the possibility of low power operation and even direct coupling into our standard SCO. If the latter can be accomplished, it will mean virtually

no additional circuitry for transducer signal conditioning. We plan to study the device in detail immediately after completion of the fabrication of the first Mark V, four-channel unit.

2.4 ECG SENSORS

Based on the experience of other workers and particularly that of G. Folk[2-3], we have decided to use stainless steel loops as the implanted electrodes.

Electrode leads will be made up from helically wound multistrand stainless wire similar to that developed for use with pacemakers. Each lead will be treated in the manner described for the thermistor sensors under Section 2.2 above.

We anticipate no difficulty with tissue reaction for these electrodes when they are used in chronic studies.

2.5 BLOOD FLOW SENSOR

We learned, during the Stockholm Meetings (1967) that our colleagues at NYU were involved in the development of an implantable blood flow sensor. We got together during 1968 because of our mutual interests. During the year we cooperated in the preparation of a telemetric breadboard able to detect the output of this sensor *ibid.* Of extreme interest is the fact that the NYU blood flow transducer *requires no external power* for its operation. It is powered by its own permanent magnet. In its present form, it produces between 1/2 to 1 μv/ml/min. of signal. This level of dc signal is extremely difficult to condition at the power levels of direct interest to us. However, we have been able to suggest new magnetic materials which, at least theoretically, show some promise for a 5 to 10X improvement in output. Of additional significance is our belief that the magnetic circuit design may be further improved with a resulting 2 to 5X improvement in output. If these levels prove to be attainable, a most promising transducer may be forthcoming.

Additional problems with the device are related to the voltage sensing electrodes. In this regard, our work with the dc potential

reference electrodes may prove directly applicable. See Section 2.6.

In its present form, the blood flow transducer weighs ca. 20 grams unencapsulated. The unit designed for a 0.5 cm lumen weighs 12 grams.

"No cost" cooperation between our two research groups will continue insofar as possible. There is no question but that our interests are complementary.

2.6 REFERENCE AND pH ELECTRODES

Some of the problems facing those interested in finding out what goes on inside unrestrained living animals are those related to the measurement of chemical and electrical activity at various sites in the animal's body. A measure of the activity is required, and this measure must be transferred outside of the animal without disturbing the animal. The transfer is accomplished by telemetry. Further, it is valuable if several variables are transmitted simultaneously so that correlation or comparison may be made between them.

Electrodes with suitable characteristics are required to transduce the status of various parameters into electrical signals in a single-valued way with certain desired precision and accuracy and without distortion or disturbance to the existing situation. These electrical signals must be of the proper magnitude, frequency response and impedance to activate the telemetry circuits.

Some of the chemical activities of interest are:

pH	(Hydrogen ion concentration)
[Na+]	Sodium ion concentration
[Z±]	in general, specific ion concentrations
ρ	conductivity of tissue, skin or internal (from chemical viewpoint)

Some of the electrical activities of interest are:

dc	potentials existing between various parts of the CNS
ρ	conductivity of tissue, skin or internal (from physical viewpoint)

EKG both rate and waveform from various sites

EEG from various sites in the brain

EMG myo-potentials from sites such as peristaltic muscles, jaw, eyelids, arm and leg

dc and ac potentials arising from blood flow directly or electromagnetically generated by blood flow from EM flowmeters

Each of these activities generates, with the proper electrodes, voltages or currents ranging approximately from 10^{-6} volts to 10^{-2} volts and/or currents from 10^{-10} amperes to 10^{-3} amperes. In general it is necessary to have a complete electric circuit in order to use two electrodes. One of the electrodes varies its potential as a function of the parameter of interest when that activity is chemical; the other does not vary. The second or "constant" potential electrode is known as the *reference electrode*. When the purpose of the electrodes is to detect an existing potential between two sites in the body, again two electrodes are required, neither of which must change with chemical activity at the site nor, if possible, for any other reason not related to a potential change. In this case, two "reference" electrodes would be called for.

2.6.1 The Reference Electrode

Silver-silver chloride electrodes properly made and aged or stabilized are about the closest which anyone has been able to come to a non-polarizable reference electrode. When used in pairs they should be placed in the same electrolyte composition so as to avoid ionic concentration voltages and further should be operated at very low current densities on the order of one microampere per square centimeter for long life. At the present time there are two "species" of silver-silver chloride electrodes, the plated and the compressed powder. In our work here in the past we have developed reliable techniques for preparing stable reproducible plated electrodes using a very high purity silver wire. These electrodes are, however, not very rugged in physiological solutions (nor are plated electrodes made by anyone else). The compressed powder electrodes are composed of a fine silver powder mixed with an approximately equal amount of silver chloride powder and other

inert additives and compressed to a solid pellet under hydraulic pressure. These electrodes are relatively rugged mechanically and because they are uniform throughout their bulk are not sensitive to scratches or deplating of the surface. However, both the plated and compressed electrodes are subject to surface contamination, surface poisoning and surface coating. Silver-silver chloride electrodes when exposed to body fluids may become contaminated by the various protein and/or fat and/or other organic molecules existing in the body fluids. These various organic materials tend with time to form a coating on the electrode surface which produces erratic action and/or drift of the electrode making the desired measurement unreliable. Various techniques to prevent this contamination are being evaluated. Coatings or coverings of cellophane and silicones have been tried with some success. In the development of a pair of electrodes to measure pH, the silver-silver chloride electrode has been chosen as the most suitable reference electrode. The calomel electrode under laboratory conditions is usually chosen as the reference electrode against a pH electrode, but the complications in its construction essentially rule it out as suitable for miniaturization. The size electrode which is considered suitable for implantation should not be larger than a few millimeters in any dimension. For this reason considerable effort has been devoted to the silver-silver chloride electrode.

2.6.2 The pH Electrode

The pH electrode measures a specific ion concentration, the universal hydrogen ion. For the application intended here (miniaturized implantable telemetry), the electrode should be insensitive to other ion species concentration, miniature (a few millimeters or less in any dimension) rapid in response, not too high in impedance, rugged, stable over a long period of time (up to a year), non-toxic to tissue and of fairly simple construction. Antimony has been used as a pH electrode and meets some of these qualifications but rapidly (a few hours) becomes unstable. In addition to the need for frequent recalibration (and for reconditioning by scraping the surface) is the large change in response due to oxygen tension which is usual in living systems. Further it is sensitive to

both reducing and oxidizing agents, carboxylic and amino acids all of which are normally present in tissues. It is completely unsuitable for long term implantation. The conductive glass electrode meets many of the above qualifications but is somewhat difficult to construct in a miniature form and does become unstable with time apparently due to a coating of protein material over its surface. Several large manufacturers of electrodes have been putting a good deal of effort into improving and miniaturizing their conductive glass electrodes, and this avenue is not closed. They are also looking into the problem of the contaminating coatings and are attempting to solve this problem by means of protective coatings.

We have been pursuing a different course. It has been reported (Ives and Janz, 1961[2-4]) and a patent (Perly, 1947[2-5]) has been issued for an iridium electrode for pH measurements. It is also suggested that rhodium, and perhaps ruthenium and osmium may be used in the same way. The characteristics of this electrode as reported from both sources are as follows: It is said to give a near theoretical response over a wide temperature range from 0 to 14 pH, and

not to be sensitive to:

carbon dioxide	copper ions
sulfur dioxide	silver ions
ammonia	lithium ions
hydrogen sulfide	sodium ions
oxidizing agents and oxygen	potassium ions
and unaffected by stirring	

to be sensitive to:

hydrogen
ionic strength

This electrode would be expected to be of low impedance, but is apparently easily polarized by current flow and hence must be used only with a very high input impedance measuring circuit. This behavior is inexplicable as the electrode is supposedly reversible. It is reported that currents greater than one-tenth microampere per square centimeter will polarize this electrode. For a wire electrode one millimeter in

diameter by three millimeters long (0.1 sq. cm.) this would require an input impedance of greater than 100 megohms. While this value can be achieved easily in laboratory instruments, it may prove difficult in miniature implantable telemeters. However, the new field-effect transistors show promise of delivering input resistances of just this order of magnitude, and since small glass electrodes may require a similar input resistance, this is not considered a serious drawback at this time.

The preparation of this electrode is somewhat special. It is recommended that it be prepared by a plating procedure which involves many careful steps of cleaning, plating from a specially prepared solution, burnishing and operating of these various steps in order to assure a bright nonporous, uniform iridium deposit with no occluded hydrogen. It appears that occluded hydrogen makes the iridium electrode inoperative and that commercial iridium is prepared in such a way that hydrogen is occluded in the solid metal. Nevertheless, it was felt worthwhile to investigate the possibility of using a spectroscopically pure iridium metal wire commercially available.

2.6.3 Preparation of Electrodes

A foot of 99.9% pure iridium wire in two sizes, 5 mil and 10 mil, was purchased from Alfa Inorganics, Inc. This was the highest purity iridium wire available. Several short lengths (1 cm) were mounted in glass tubes to form electrodes both by sealing directly to the glass and by means of the best epoxy found for this purpose (see electrode sealant tests). Approximately 5 millimeters of the wire extended beyond the glass lip. Connection was made inside the glass tube by welding or soldering the necessary lead wire to the iridium.

The configuration shown in Figure 2.6-1, or one similar to it, was used for all electrode tests as it was simple, inexpensive and rapidly made. When cement seals were made the glass was preshrunk at the tip to form a snug fit to the wire. This allowed a minimum amount of cement to be used in the seal and exposed the smallest possible amount of cement to the test solutions. Several iridium electrodes were tested against a

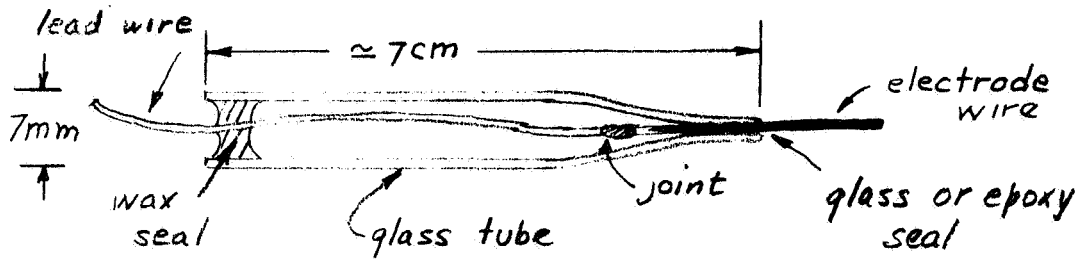

Figure 2.6-1. Test Electrode Configuration

standard Beckman mercury-mercurous chloride (calomel) reference electrode in various standard buffer solutions (pH 5.58, pH 7.0 and pH 8.48 ± pH 0.2 @ 25°C), Harleco Reference buffer solutions, Beckman Buffer pH 6.86 and A. H. Thomas pH 4.01 ±.01, pH 7.00 ± .01 and pH 10.00 ± .01.

The pH was read by means of the Beckman Expandomatic pH meter. This meter has a long scale 0 to 14 pH or can be expanded to read any chosen 2 pH over the full scale. Voltage may be directly read between electrodes as millivolts 0-1400 mv or any 200 mv span within the range. Accuracy (relative to buffer standard) is ± .05 pH standard scale, ± .01 expanded scale. Manual temperature compensation is provided. This meter utilized negative feedback circuits to provide extremely high input resistance (so that electrodes will not be loaded and polarized). The input impedance is not given for this meter but is believed to be greater than 100 megohms. However, difficulty was found in trying to measure the pH of distilled water presumably due to the high resistance of the water (resistance in standard conductivity cell ≃ 3 megohms).

Measurements were also made using the Keithley 610A Electrometer. This instrument can measure to 10 millivolts full scale upward at any desired input resistance from one ohm to 10^{14} ohms in factors of ten. This is not useful in determining polarization current since at, for example, one volt, the current drawn can be adjusted anywhere in the range

from one ampere (of course the source cannot sustain this) to 1/100 of a picoampere (1/100 picoampere = 10^{-14} ampere). One difficulty with this instrument is that considerable drift occurs in the instrument itself and this makes it quite difficult to separate instrument drift from electrode drift.

Some early measurements made on iridium electrodes against standard calomel electrodes are shown in the figure below.

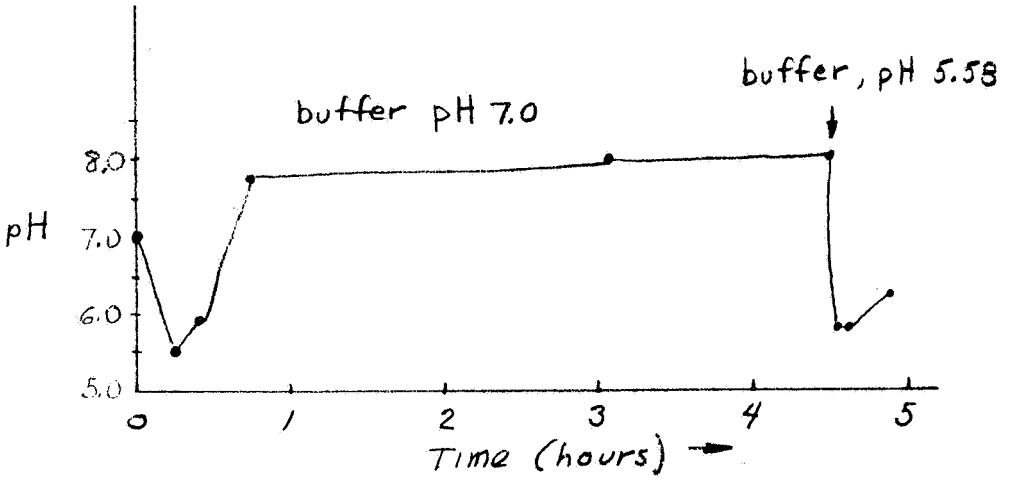

Figure 2.6-2. Iridium Electrode Against Calomel, Beckman pH Meter, Iridium Scraped and Cleaned in NHO_3 Ir Electrode Output vs. Time

This indicates a large initial drift, return and finally a slow upward drift. The initial change may have been due to the HNO_3 treatment. Later tests did not use the acid cleaning.

Tests were made changing the electrodes rapidly from pH 7.0 to pH 5.58 and back again. When this was done the correct difference of 1.42 (or nearly so) pH always occurred although the absolute reading of pH gradually drifted upward at both the high and low pH values to keep the difference approximately constant.

Later measurements using the Keithly electrometer gave the following data with 10^4 ohms input impedance.

0.115 volts pH 7.0

0.150 volts pH 5.6

Several minuts were required for a stable value to be reached. The same stable value was reached after removal and replacement of the electrodes.

A typical preliminary test of the iridium electrode vs. calomel gave the following curve.

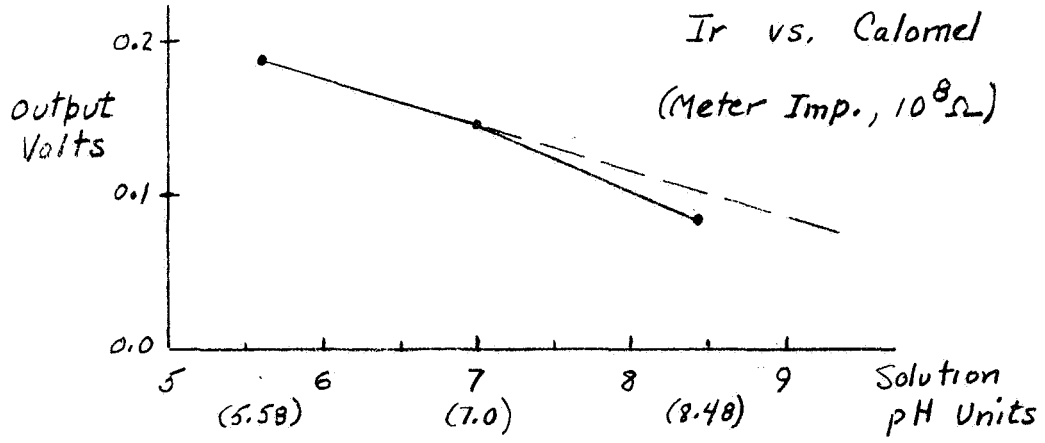

Figure 2.6-3. Iridium-Calomel Electrode Pair Voltage vs. pH (allowed to stabilize 20 minutes)

Further tests were performed using the same iridium electrode vs. a geometrically similar silver-silver chloride electrode. The relationship is shown in Figure 2.6-4.

While these curves are not linear, they show a good change in voltage vs. pH both for Ag-AgCl and calomel as the reference electrode. Further

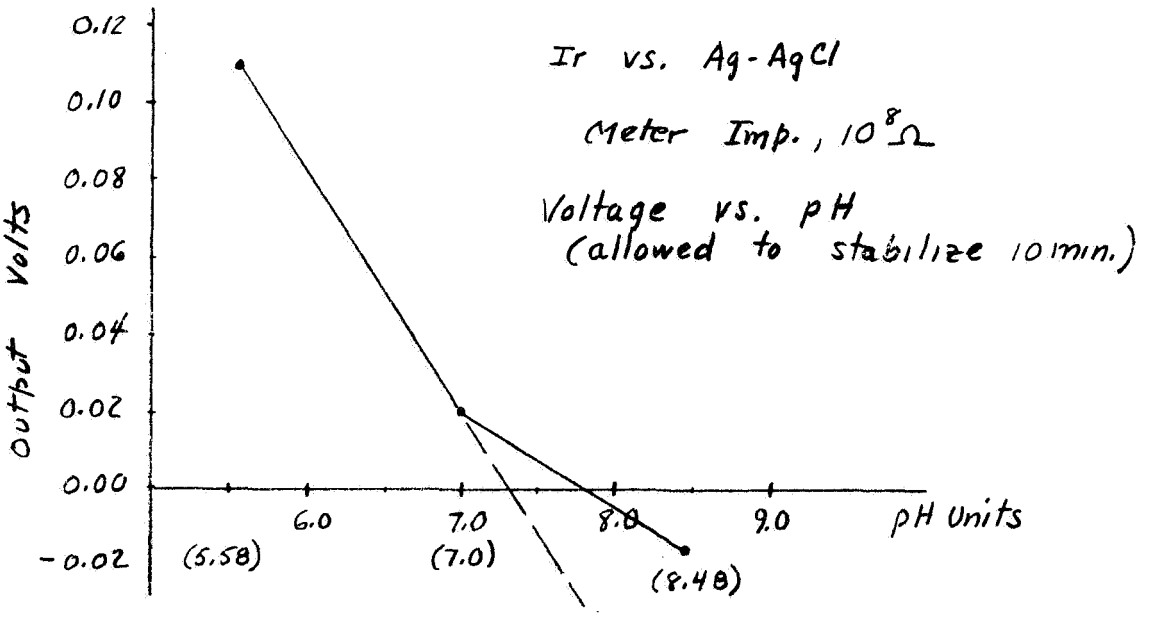

Figure 2.6-4. Output Volts vs. pH for Ir-Ag-AgCl Pair

work using these iridium electrodes is not contemplated. Plated iridium electrodes will be used for future work as it is believed that the material used for the previously described electrodes is not suitable for obtaining voltages equivalent to theoretical voltages, whereas the plated electrodes are reputed to do so. Measurements will be carried out at both about 10^8 ohms impedance and 10^{14} ohms impedance. The higher impedance values should permit correspondence more closely to theoretically expected values while those at 10^8 ohms will be performed because this is approximately the practical value of impedance that may be expected from present miniature telemetry circuits.

Silver-Silver Chloride Reference Electrodes

As was explained previously, electrodes of a stable reversible nature are required as reference electrodes against which a species sensitive electrode must work. For testing purposes a configuration was developed

which was easy to construct and rugged. The details are shown in Figure 2.6-5.

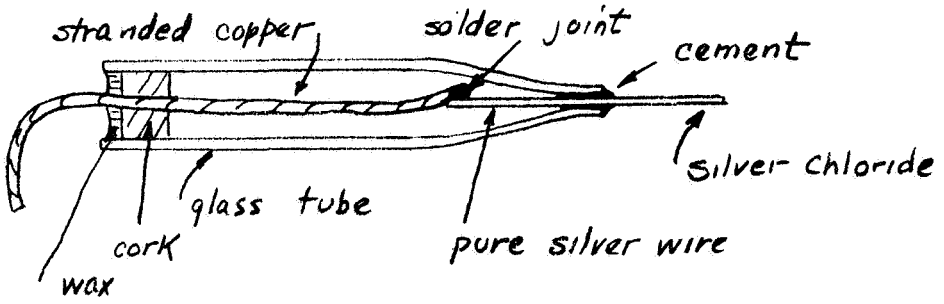

Figure 2.6-5 Silver-Silver Chloride Test Electrodes

Electrodes have been prepared in either of two ways. After cleaning the pure silver wire (99.99+ % pure) (0.05 cm dia. x 1.5 cm) in NHO_3 and distilled water until a snowy white velvet like appearance is obtained which wets uniformly, the silver wire is soldered to the stranded copper wire with a minimum of flux and solder. At all times the silver wire is handled with clean tweezers and never touched with bare fingers. The glass tube is prepared ahead of time by shrinking the tip in a Bunsen flame so that it makes a close fit to the wire. At this point the wire may be either (1) plated with silver chloride and then connected into place or connected into place and then coated with silver chloride. The latter procedure seems to be preferable. If plated before cementing, approximately 0.75 cm of the silver wire is plated. In either case only 0.5 cm of the wire is allowed to extend beyond the glass; the remainder is sealed inside thus forming a tip of 0.05 cm dia. x 0.5 cm long. The exposed electrode then has a total area of very nearly 0.1 cm^2. The plating procedure is as follows: A small beaker is used into which is inserted a thoroughly cleaned platinum cathode; 0.1 N Hcl solution is used as the electrolyte. It has been recommended[2-4] that a well used and/or thoroughly aged electrolyte solution forms a

better silver chloride coating than a fresh solution. This has been found generally to be true, and after use the solution is returned to stock. Solutions several years old are continuing to give excellent results. Although current densities of up to 18 ma cm^{-2} have been recommended[2-4] as producing good silver chloride coatings, we have found that the best, most uniform, coherent and pore-free layers are formed at a lower current density. We have consistently obtained the best results at a current density of 1.0 ma cm^{-2}. Plating is done for 30 minutes, producing a plum brown color coating. After plating, the electrode is thoroughly rinsed with distilled water, soaked in several changes of distilled water for about a day, dried and stored in a low humidity cabinet. The thorough washing removes any free chloride from the silver chloride matrix. Electrodes made by this procedure have proved to be uniform in appearance and most important, to be low and similar in interelectrode residual potential. Generally these electrodes showed less than one millivolt maximum potential difference between any pair in a set of ten in 0.154 N saline when new.

After stabilizing by shorting together and immersing in 0.154 N. saline overnight, the maximum difference was less than 0.1 millivolt. Pairs could be selected to have negligible potential difference of less than 50 microvolts difference. With great care in all steps of cleaning, plating and washing, even greater uniformity can be obtained.

Electrode Coating

One of the important requirements for a long-life implantable electrode is a means of protection of the electrode so that it will maintain its original physical characteristics upon which its electrical characteristics depend. This protection must not change or inhibit its electrical characteristics, however. Various techniques have been tried to achieve this. Cellophane and various gells have been used with some success. In the course of our investigations of various encapsulant materials, we have found that Sylgard* while poor as a protection for

*Dow Corning #184 Potting Resin, also variations by Dow Corning and a similar product by General Electric

implantable electronic circuits because of its high vapor permeability, for this same reason may be useful as a membrane for electrode protection.

A series of silver-silver chloride electrodes were constructed as described previously and the tips were then coated with Sylgard resin. These electrodes performed identically in various buffer solutions to the uncoated ones, Figure 2.6-6.

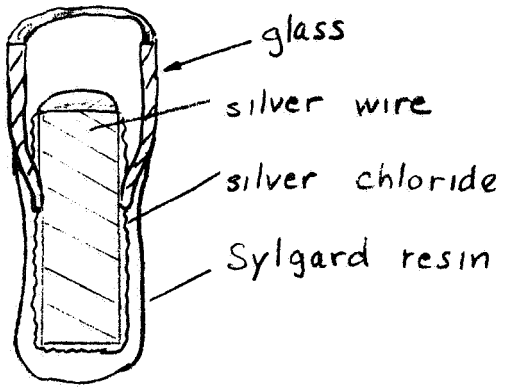

Figure 2.6-6 Enlarged View of Coated Electrode

Apparently the permeability to water molecules and ions was sufficiently high to allow the electrode to perform properly. Even a heavy coating achieved by several dipping and curing cycles seemed only to slow down the response time from almost instantaneous to about 15 to 30 seconds. Another configuration, Figure 2.6-7, which should be more rugged, will be tried shortly. Since the Sylgard does not adhere strongly after long-time immersion, the improved geometry of this configuration should make the whole assembly more rugged.

Silver-silver electrodes are concentration sensitive, that is, the half cell potential is a function of the chloride ion concentration in which they are immersed. In order to reduce this sensitivity the electrode can be immersed in a fixed concentration of chloride ion, say 0.1 N KCl, and connected by a bridge of this solution to the desired solution. In an attempt to achieve this condition essentially as an integral part of the electrode, the configuration shown in Figure 2.6-8 was tried.

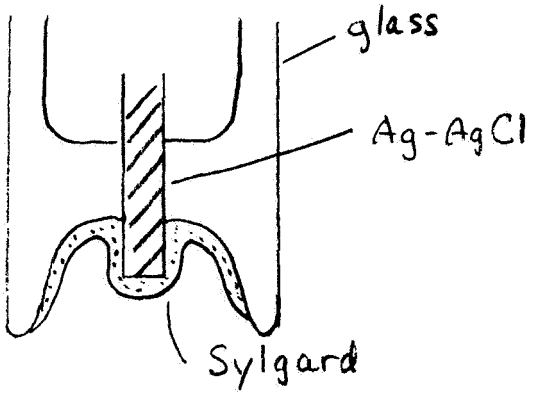

Figure 2.6-7. Future Design for Membrane Coated Electrode

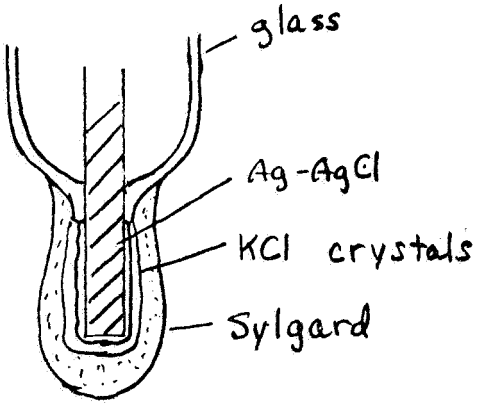

Figure 2.6-8. Ag-AgCl/KCl Electrode

It was hoped that the KCl crystals would, when the electrode was immersed in the liquid to be measured, form a long-lasting saturated KCl solution in intimate contact with the AgCl. This solution would then, in effect, stabilize the junction potential and minimize sensitivity to chloride concentration of the measured electrolyte. From some preliminary measurements made on a few electrodes of this type, it is believed that they operated as expected. However, several difficulties were encountered with these electrodes. First the KCl crystals quickly dissolved in a few hours. Upon examination of the electrodes, it was found that the Sylgard had disintegrated in places and no longer formed a continuous film, thus the KCl simply dissolved into the test solutions. In addition, the silver chloride plating was dissolved in spots exposing the underlying silver wire. This is believed due to the high concentration of KCl at the point of contact with the silver chloride. In general then, the attempt to form a self-contained intermediate KCl solution between the electrode surface and the covering membrane was not unsuccessful. While it would be desirable, but not absolutely necessary, to have this intermediate electrolyte, this problem will not be given a high priority, although a few more electrodes of this type will be made in case a simple solution to the problem can be found. It is felt that the use of a Sylgard membrane, without the KCl crystals, is successful and that it will prove to be quite useful.

While experimenting with the Sylgard resin, attempts were made to form thin films on a water surface. It was found that a critical variable in the curing was the pH of the water. Films made on pure distilled water did not cure properly. By the addition of a few drops of HCl the water was made acidic (pH $\simeq$ 3.0) and then the Sylgard formed tough, fully cured films. If the water was basic (even approximately 8 pH), the films would not cure at all and remained fluid and/or gummy.

Several dozen thin films on the order of 0.0005 to 0.005 inches were made in this way and appeared uniform in thickness and hole free except at the edges. It is felt that these films might be useful for protecting electrode surfaces and for use as permeable membranes between electrolytes in electrodes requiring this.

One silver-silver chloride-saturated KCl electrode was prepared which performed exactly as predicted. This electrode incorporated an asbestos fiber junction. The construction is shown in Figure 2.6-9.

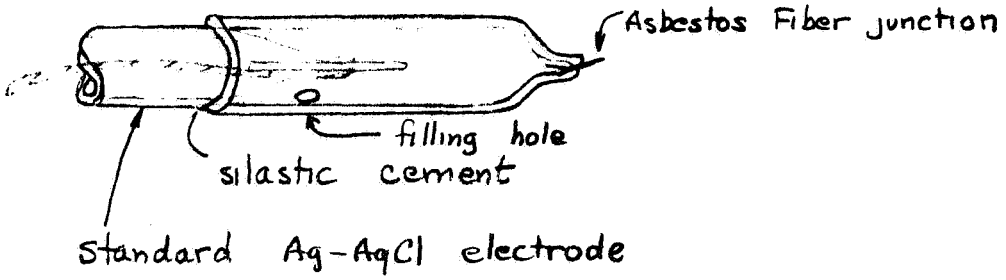

Figure 2.6-9. Ag-AgCl-KCl Electrode with Asbestos Fiber

The lower chamber is filled with KCl of a known concentration: 0.1 N, 1.0 N, or saturated, to a level to contact the AgCl electrode. The asbestos leak then allows a minute quantity of the solution ($\simeq 10^{-7}$ cc sec.) to escape from the reservior, thus keeping a constant concentration difference between the two sides of the junction. The chief difficulty of this arrangement is that it cannot be used in an implant for two reasons. One, it depends on gravity to provide the force to maintain the flow through the junction, and, two, the reservior has a finite capacity and thus limited life. In addition, junctions of this type are highly susceptible to clogging and thus cease to provide a conducting path for the electrical current required.

Cements and Adhesives

A number of cements, sealants and adhesives were tested for use in the preparation of electrodes. They ranged through many classes of cements and through a number in any class. Several epoxy cements, natural resins, cellulose, vinyl, etc. were tested. The epoxies were tested at various cure cycles and with different proportions of

resin-to-catalyst ratios. The best ratios are reported in Table 2-1. Various methods of testing were considered, and a test was devised which approximated the actual application. It was noticed in preliminary testing that, in general, the cement itself did not deteriorate due to immersion in water, but rather that a thin film of water would creep in from the edges, eventually lifting the cement from the surface. The following test was then devised. On a clean sheet of glass (microscope slide) approximately 50 drops of cement, 2 to 3 mm in diameter, were placed. The glass was permanently marked as to the kind of cement, curing, etc. These drops of cement were then cured according to the desired procedure for this kind of cement. After curing, the sheets were placed in a beaker of water and allowed to remain for 25 hours. At the end of this time the sheet was removed and dried by absorbent paper, and the adhesion of a few of the spots were tested by means of a needle. The needle was pressed on the spot at the glass to cement junction. Good spots required a great deal of pressure to cause them to pop off whole. Excellent spots would break off in pieces. Poor drops could be wiped off by the absorbent paper. A rough scale was used from 1 through 10 as to the goodness of adhesion; this is explained in the table.

TABLE 2-1

	Class of Cement	Number and Manufacturer	Comments—Preparation and Cure	Evaluation	Rating*
1.	Epoxy	0151 Hysol Clear	Resin to Hardener ratio 3:1 best, 1 hour 60° C or 24 hrs. room temp.	one of better ones	8-9
2.	Epoxy	1C Hysol White	4:1 best to 6:1 best	better than 0151, clear spots break up apply to cold surface	8-9
3.	Epoxy	E-63 Techkits	mix as recommended	poor	1-2
		A-12 Techkits	mix as recommended 1 hour 70°C		5-6
		E-7 Techkits	mix as recommended	E-7 softens in $H2_0$	6
4.	Epoxy	24 Eccobond Emmerson & Cummings	best ratio 4:1	Not particularly good	2-4
5.	Epoxy	Scotch Structural Adhesive -3M	Baked or unbaked best ratio 3:2	soft, spots came off whole	2-3
6.	Epoxy	304 Chem-Lok	as recommended (spongy)	whole spots come off, soft	4
7.	Epoxy	201 Epotec Epoxy Technolgy Inc.	1:1 best - clear 10 minute cure	best of all tested, cure 24 hours adheres well after months immersion	9-10
8.	Epoxy	Homelite Homelite Co.	clear, wide range hardener, ratio used- 8:1 best	Inferior to 1, 2, 7	6-7
9.	Silicone	630 Protective coating, Dow-Corning, solvent evap., silicone wax	dip-solvent evap.	almost as good as paraffin to moisture resistance, will not melt, very low mechanical strength, waxy	9

	Class of Cement	Number and Manufacturer	Comments--Preparation and Cure	Evaluation	Rating*
10.	Silicone rubber	Glass and ceramic cement, Dow-Corning	Absorbs moisture from air to cure - 24 hours cure, but difficult to manipulate - gummy	Strong, excellent	9
11.	Sylgard	No. 184 Potting Resin - Dow-Corning	Flows and casts well, cures 24 hours, room temperature; 2 hours @ 100°C, rubberlike	Tears medium strength, water permeable	2
12.	Polyethylene	Low molecular weight	Thermoplastic, difficult to make flow without burning	some good some excellent	7-8
13.	Ceramic (Silicate)	S-29 Sauereisn	Handles well, hardens 24 hours	Rock hard dry cannot be separated from glass dry, softens in H_2O , depends on bake	2-6
		S-33	Grind to smaller particle size, handles well, hardens 24 hours	poor	2-6
14.	Gum	Damar	Solvent-xylene solvent dry	Long time for large spots to dry, hard to gummy	5
15.	Gum	Canada balsam	Solvent evaporates	Chipped off readily after immersion	4
16.	Wax	Apiezon "W"	Thermoplastic, apply to hot surface	Good until immersed	2-3
17.	Paraffin	Many kinds	Apply to warm surface	Very weak mechanically	7
18.	Paraffin with vinyl copolymer	Many kinds	Apply to warm surface	Stronger than paraffin alone	7
19.	Shellac	Fisher By-seal Hi-temp Moderate temp Low temp.	apply to hot surface	Very brittle poor in small spots	1

	Class of Cement	Number and Manufacturer	Comments-Preparation and Cure	Evaluation	Rating*
20.	Bakelite	Bakelite cement, General Cement Company	Air dry and bake 1 hour, 100°C	Brittle	4-6
21.	Glytol	G. E. Glyptol paint	Air dry or bake or apply to hot surface	Brittle	3-4
22.	Cellulose	General Cement	Air dry or bake	poor	1
		Duco			11
		Everfast			1
		Weldit		best of cellulose	4
23.	Vinyl	Dab-Hollingshead	Air dry	water softened	1

*Rating Semiqualitative (1 through 10) as to goodness of stick to glass slide after 3 days immersion in water. Three mm. dia. drops tested with needle point

Scale

1 Very poor - most spots slipped off by wiping
2 Very poor, some spots slipped off by wiping
3 Poor - touch with needle would remove spot
4. Poor - some force required to remove spots whole
5. Medium - medium force required to remove spot
6. Medium - some spots broke, some off whole, medium force
7. Good - many spots broke before coming off - above medium force
8. Good - most spots broke before coming off - above medium force
9. Very good - high force required to break or remove spot - prolonged immersion
10. Excellent - very high force to break spot - withstood prolonged immersion

Mixes - Where resins and hardener were mixed, many ratios were visually tested--the best is reported--this is not necessarily the same as recommended by manufacturer.

Curing - Cure of resins was at least twice the time recommended by manufacturer at room temperature. When cured at elevated temperature, recommended times and temperature used otherwise as noted.

Surface Preparation - New, precleaned glass microscope slides were used for all tests. These were wiped with absorbant paper saturated with pure alcohol and wiped dry. Some tests were performed with heating to > 200°C in Bunsen flame in addition to above. No improvement seen by heating.

The best cements as determined by the above evaluation are, in order: EPOTEC No. 201 (1:1), Hysol 1-C (5:1), Hysol D151 (3:1). These cements when applied to a clean surface and given a proper curing cycle will adhere on the order of months under water immersion. The clear resins (i.e. without filler) will probably add less impurities to the electrode structure, although no deleterious effect has been noted from the Hysol 1-C filler, which is probably an inert metallic oxide or silicon dioxide. One or more of these three will probably be used for future experimental and implantable electrodes. Surface roughness has not been investigated with respect to holding ability, but some tests will be run shortly comparing polished glass with etched or abraded surfaces.

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3.0 COMMUNICATIONS AND COOPERATION

3.1 COMMUNICATIONS

During the period being reported, we were visited by a number of researchers and communicated by mail with many others.

Early in the period Mr. Heinz Wolff of the Holly Hill Laboratory (Medical Research Council), London, visited with us. We discussed our SCO designs and multichannel implant development with him, as well as other work involving the measurement of intracranial pressure measurement. We were able to give him information and test data on the new mercury cells developed by the Union Carbide Corporation as well as a set of sample units.

Dr. Ronald Barr from the University of Missouri returned the units supplied him for irradiation. These were standard Mark IV-LL units. They were irradiated to the following accumulated levels:

SN401: 280.75 Krads
SN402: 480. Krads (in saline)
SN403: 100. Krads (in saline)

Of these units, serious deterioration in performance was reported by Dr. Barr for SN's 401 and 402. No measurable change was noted on SN403 (100 Krads). However, we do not yet know if aging or other subtle changes accrued. It is still our hope to complete these studies and report results in detail, jointly with Dr. Barr.

We were visited by Dr. G. R. Hervey, of the Department of Physiology, The University of Leeds. Dr. Hervey was keenly interested in our Managed Energy Terrella (MET) studies and our approach to the concepts of Biological Experiments. We discussed our work in detail and subsequent to his visit, sent him copies of our report sections in which he expressed interest.

We have been in frequent communication with the staff of the Hyperbaric Facility and Vascular Surgery at the Institute for Rehabilitation Medicine of the New York University Medical Center. They are deeply interested in relating NASA-supported multichannel implant developments to their studies of the cardiovascular system function. We are interested in their on-going developments in permanent magnetic blood-flow transducers and diffused piezo-resistive pressure sensors. Our cooperation is expected to result in implant evaluations productive of biological and unit performance data of direct interest to the Bioscience Programs effort.

We have communicated with Dr. John Thach of the Naval Aerospace Medical Center, NAMI, Pensacola, Florida. He has been particularly interested in the possibility for application of combined Mark IV and Mark V implant device techniques for use with primates in space related studies.

Drs. Stevenson and Sackett of the University of Wisconsin Primate Laboratory have been interested in problems somewhat similar to those of Thach, but for ground-based studies. We have been in active communication with them in this regard.

We have been able to apply Mark IV implant techniques for a specific forest population dynamics study in Canada. The study is under the direction of Dr. Gerald G. Marten of the Petawawa Forest Experiment Station (Ontario).

From time to time throughout the period we have been in communication with Dr. Franz Halberg and Dr. Walter Ruenge of the University of Minnesota. Information on a variety of Mark IV fabrication and encapsulation techniques has been communicated to Dr. Ruenge. Data on sources of specialized material have also been given. Dr. Halberg's facility was given a Mark IV-PLLA unit for evaluation purposes as was the Biosatellite Project at the Ames Research Center. These units were designed as long-life miniature Mark IV implants with improved spectral and drift characteristics.

Mark IV devices and techniques for application were communicated to Dr. Robert C. Bolles, Department of Psychology, University of Washington, Seattle. We understand that he has realized excellent performance and data in his application.

Communications were initiated with Dr. Carl Gans at the State University of New York at Buffalo based on information from Dr. Jacobs (NASA). We expect a personal visit by Dr. Gans later in the year.

Prior to the first Biosatellite (II) flight we had the pleasant opportunity to meet Dr. and Mrs. Charles Lyon of Dartmouth College, Hanover, New Hampshire. At that time Dr. Lyon conducted a brief ground study of wheat seedling development in a simple magnetic shield. Much more recently we have been in further mutual communication, and we look forward to the opportunity to make available to him techniques and data developed in MET-related studies.

During the period we were host to the BIAC group and took the opportunity to outline our work in implantable devices, null magnetic field studies and bio-systems analysis.

3.2 PAPERS AND MEETINGS

A paper (2-1) (ibid) was coauthored by R. J. Gibson, Jr., R. M. Goodman (both of FIRL) and T. Reich and M. Youdin, both of the NYU Rehabiliation Institute. It was presented by Mr. Youdin at the Annual Conference on Engineering in Medicine and Biology, Houston, Texas, in 1968. The circuitry involved is based fully on Mark V developments.

3.2.1 Papers Planned or in Preparation

A monograph on magnetics and magnetic shielding oriented for biologists and biological research is included as part of this report. It was written to bring together—in a useful manner—our experience here. It is also written for dissemination as a reference and/or tutorial monograph of interest to the biological community. It was prepared by R. J. Gibson, Jr.

A paper on dual channel telemeters using a combination of Mark IV,

Mark V and proprietary techniques is in preparation by A. Marmarou of our staff.

A paper is planned on the new pulsed fm/fm ECG/temperature unit described in this report. Once implant data are in hand, the paper will be authored by R. M. Goodman.

On completion of its fabrication and calibration, the 4-channel Mark V unit will be implanted for performance study. It will be fully reported in an article planned by R. M. Goodman.

This report has been written by:

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